US011986327B2

(12) United States Patent
Gregerson et al.

(10) Patent No.: US 11,986,327 B2
(45) Date of Patent: May 21, 2024

(54) **MEDICAL X-RAY IMAGING SYSTEMS AND METHODS**

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(*) Notice: Subject to any disclaimer, the term of this patent is extended or adjusted under 35 U.S.C. 154(b) by 372 days.

(21) Appl. No.: **17/524,201**

(22) Filed: **Nov. 11, 2021**

(65) **Prior Publication Data**

US 2022/0061780 A1 Mar. 3, 2022

Related U.S. Application Data

(63) Continuation of application No. 16/354,047, filed on Mar. 14, 2019, now Pat. No. 11,197,643.

(Continued)

(51) **Int. Cl.**
A61B 6/03 (2006.01)
A61B 6/00 (2006.01)
(Continued)

(52) **U.S. Cl.**
CPC ***A61B 6/035*** (2013.01); ***A61B 6/4405*** (2013.01); ***A61B 6/4411*** (2013.01);
(Continued)

(58) **Field of Classification Search**
CPC A61B 6/035; A61B 6/4405; A61B 6/4411; A61B 6/4429; A61B 6/4447; A61B 6/54;
(Continued)

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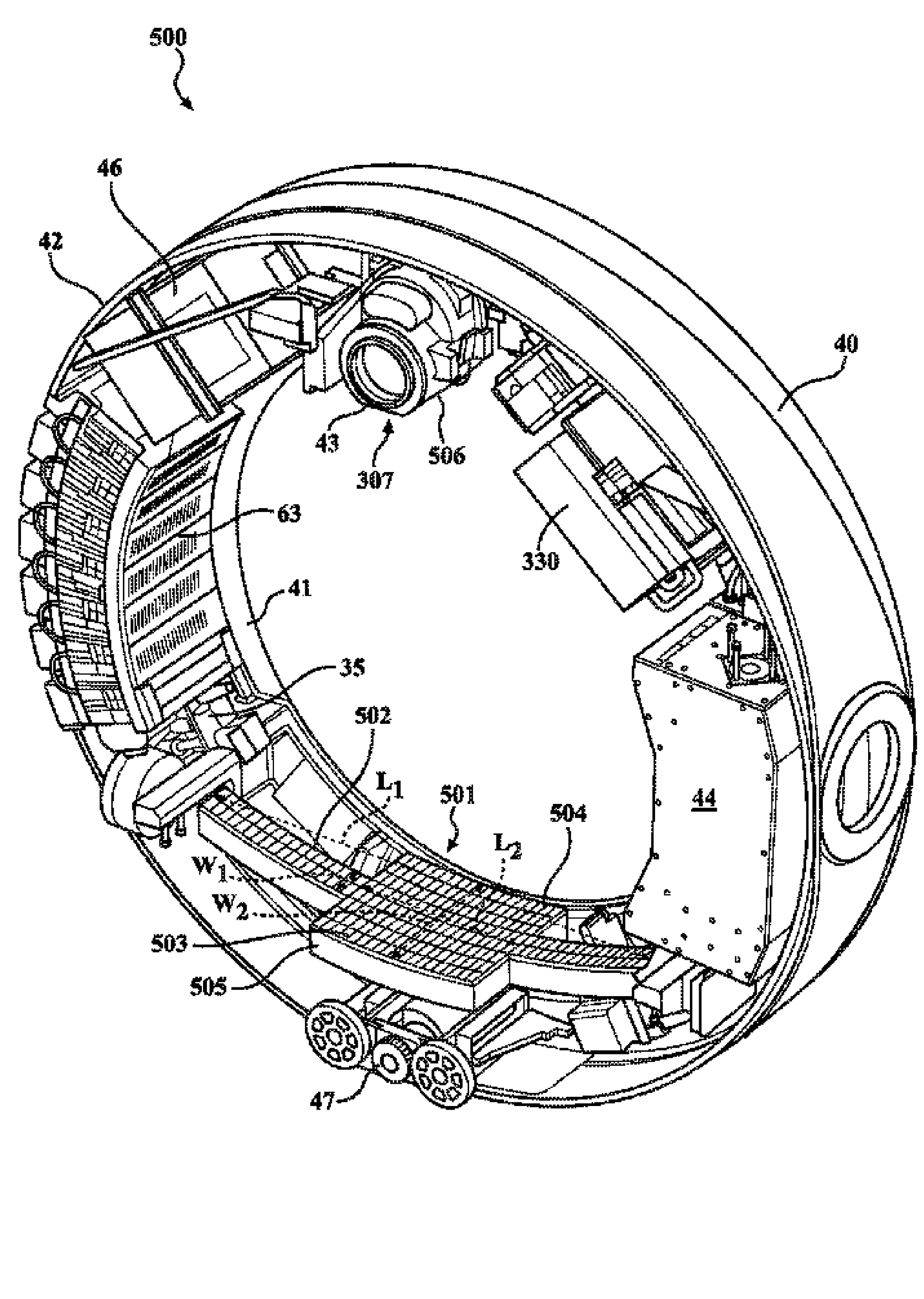
(57) **ABSTRACT**

A detector system for an x-ray imaging device includes a detector chassis, a plurality of sub-assemblies mounted to the detector chassis and within an interior housing of the chassis, the sub-assemblies defining a detector surface, where each sub-assembly includes a thermally-conductive support mounted to the detector chassis, a detector module having an array of x-ray sensitive detector elements mounted to a first surface of the support, an electronics board mounted to a second surface of the support opposite the first surface, at least one electrical connector that connects the detector module to the electronics board, where the electronics board provides power to the detector module and receives digital x-ray image data from the detector module via the at least one electrical connector. Further embodiments include x-ray imaging systems, external beam radiation treatment systems having an integrated x-ray imaging system, and methods therefor.

22 Claims, 23 Drawing Sheets



Related U.S. Application Data

(60) Provisional application No. 62/644,032, filed on Mar. 16, 2018.

(51) **Int. Cl.**

A61N 5/10	(2006.01)
H01L 27/146	(2006.01)
A61B 6/42	(2024.01)
A61B 6/58	(2024.01)

(52) **U.S. Cl.**
CPC ***A61B 6/4429*** (2013.01); ***A61B 6/4447*** (2013.01); ***A61B 6/54*** (2013.01); ***A61N 5/1064*** (2013.01); ***H01L 27/14618*** (2013.01); *A61B 6/4266* (2013.01); *A61B 6/4291* (2013.01); *A61B 6/587* (2013.01); *A61B 6/588* (2013.01); *A61B 2560/0437* (2013.01); *A61B 2560/0443* (2013.01); *A61B 2562/166* (2013.01); *A61B 2562/227* (2013.01); *G01N 2223/301* (2013.01); *G01N 2223/321* (2013.01); *G01N 2223/3303* (2013.01); *G01N 2223/3304* (2013.01)

(58) **Field of Classification Search**
CPC A61B 6/4266; A61B 6/4291; A61B 6/587; A61B 6/588; A61B 2560/0437; A61B 2560/0443; A61B 2562/166; A61B 2562/227; A61N 5/1064; H01L 27/14618; G01N 2223/301; G01N 2223/321; G01N 2223/3303; G01N 2223/3304
See application file for complete search history.

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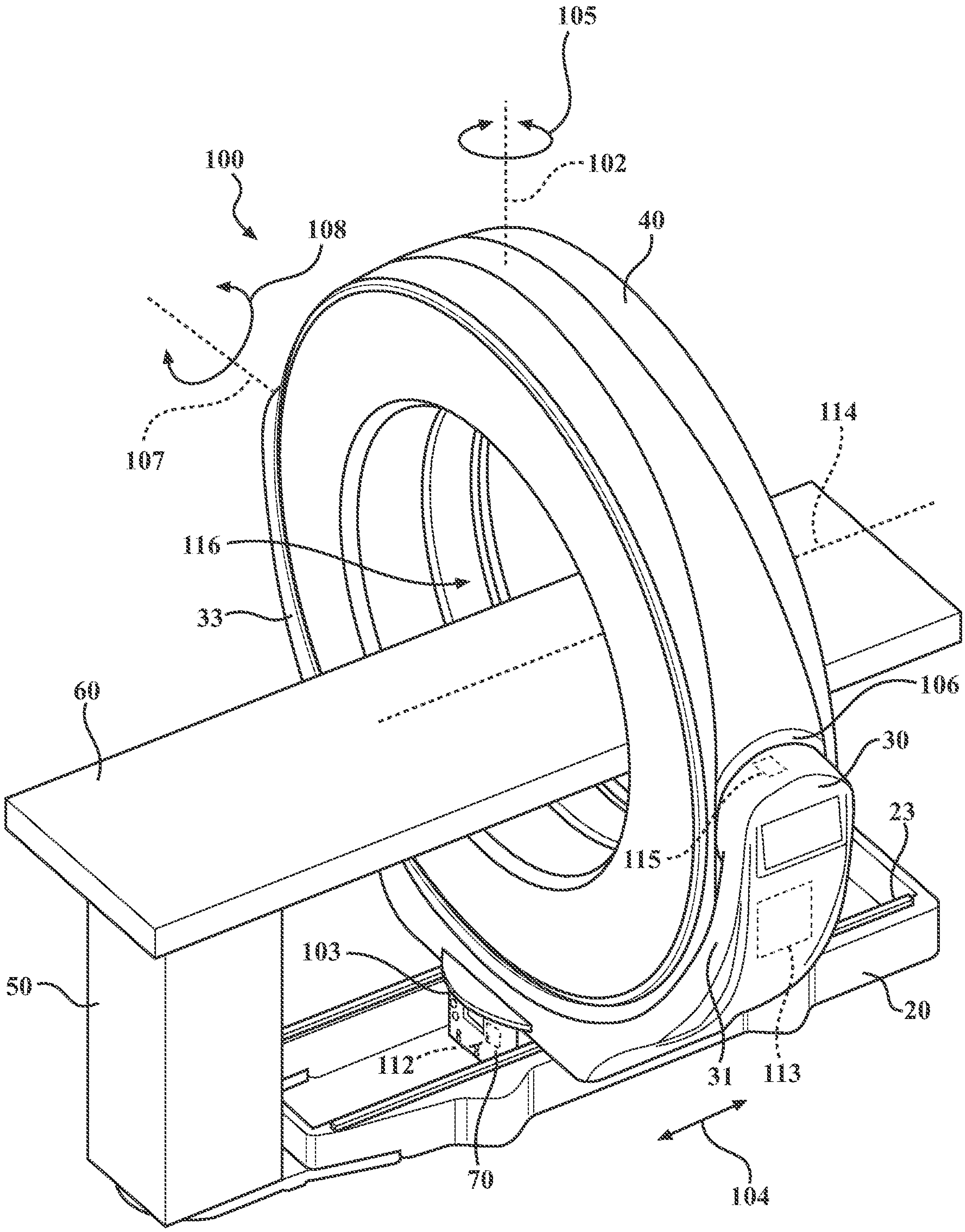
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FIG. 1

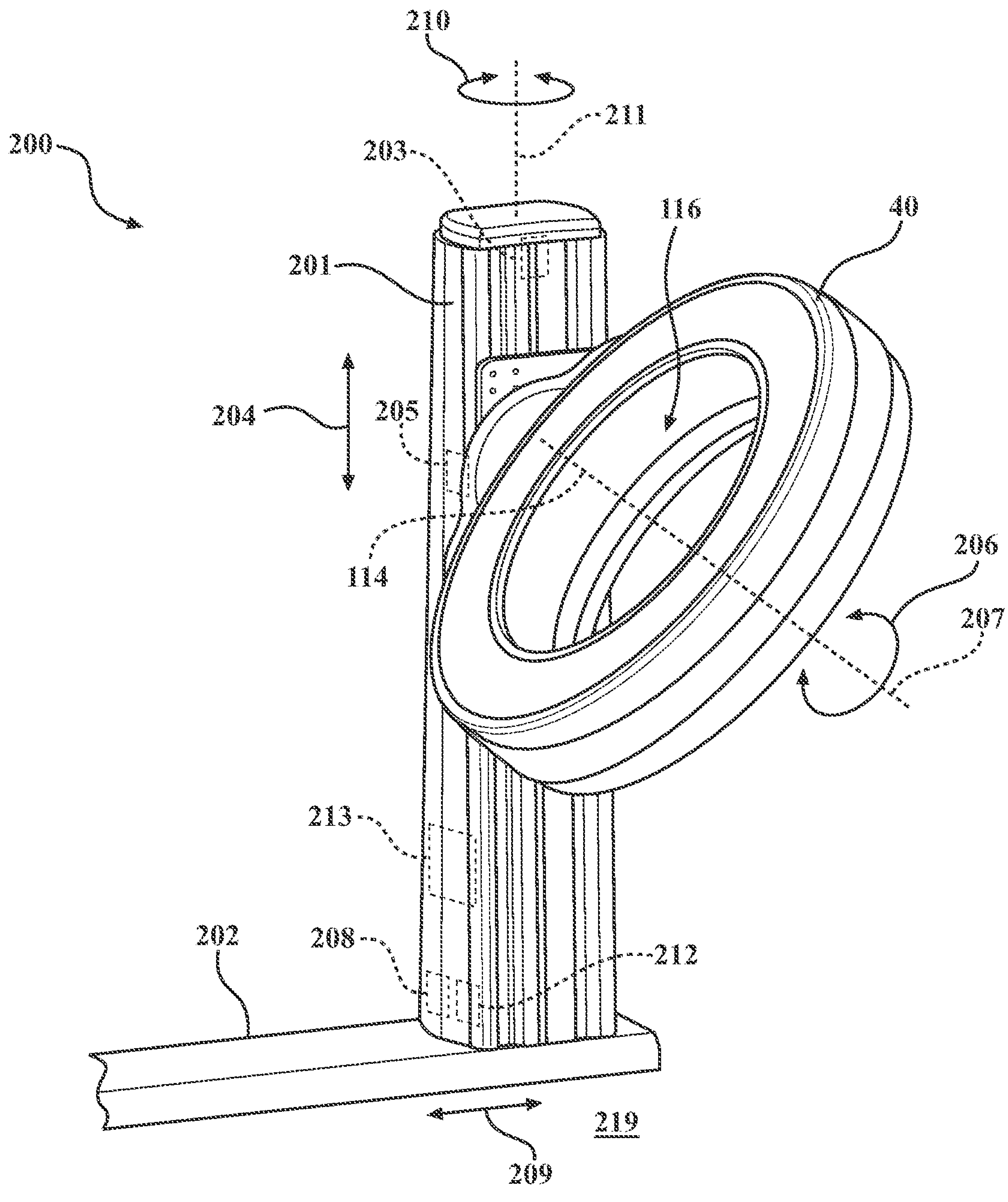
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FIG. 2

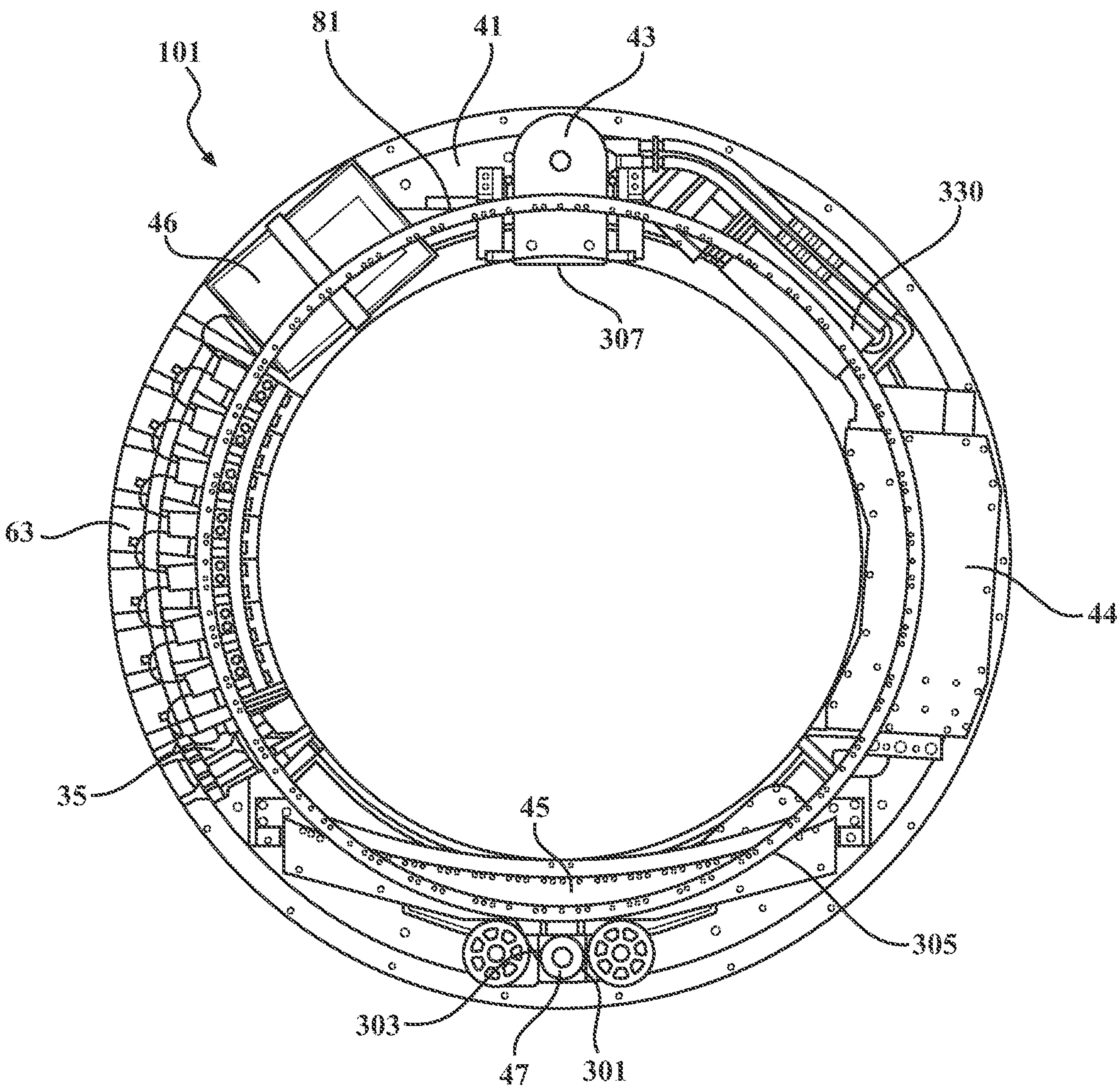

FIG. 3

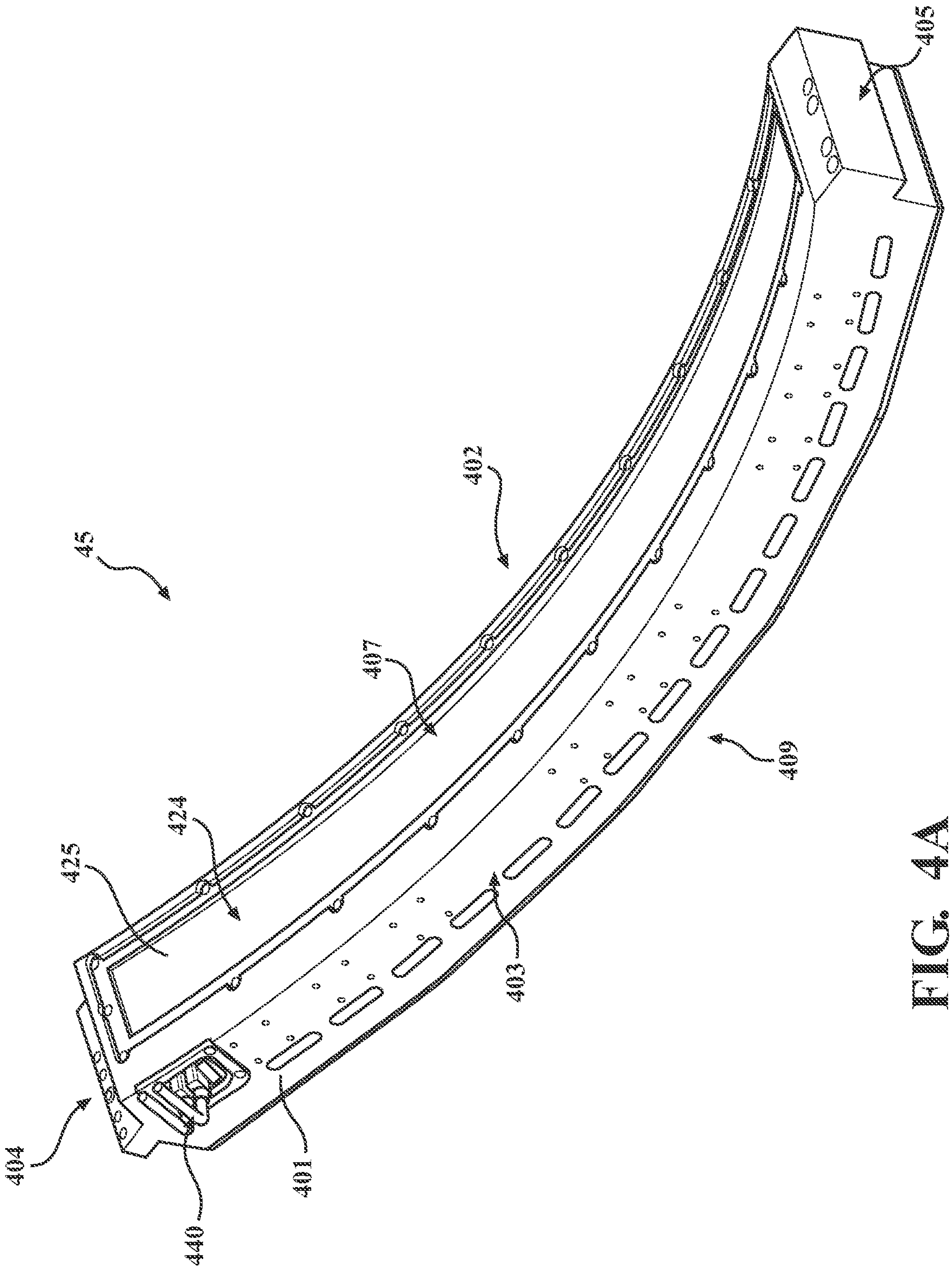

FIG. 4A

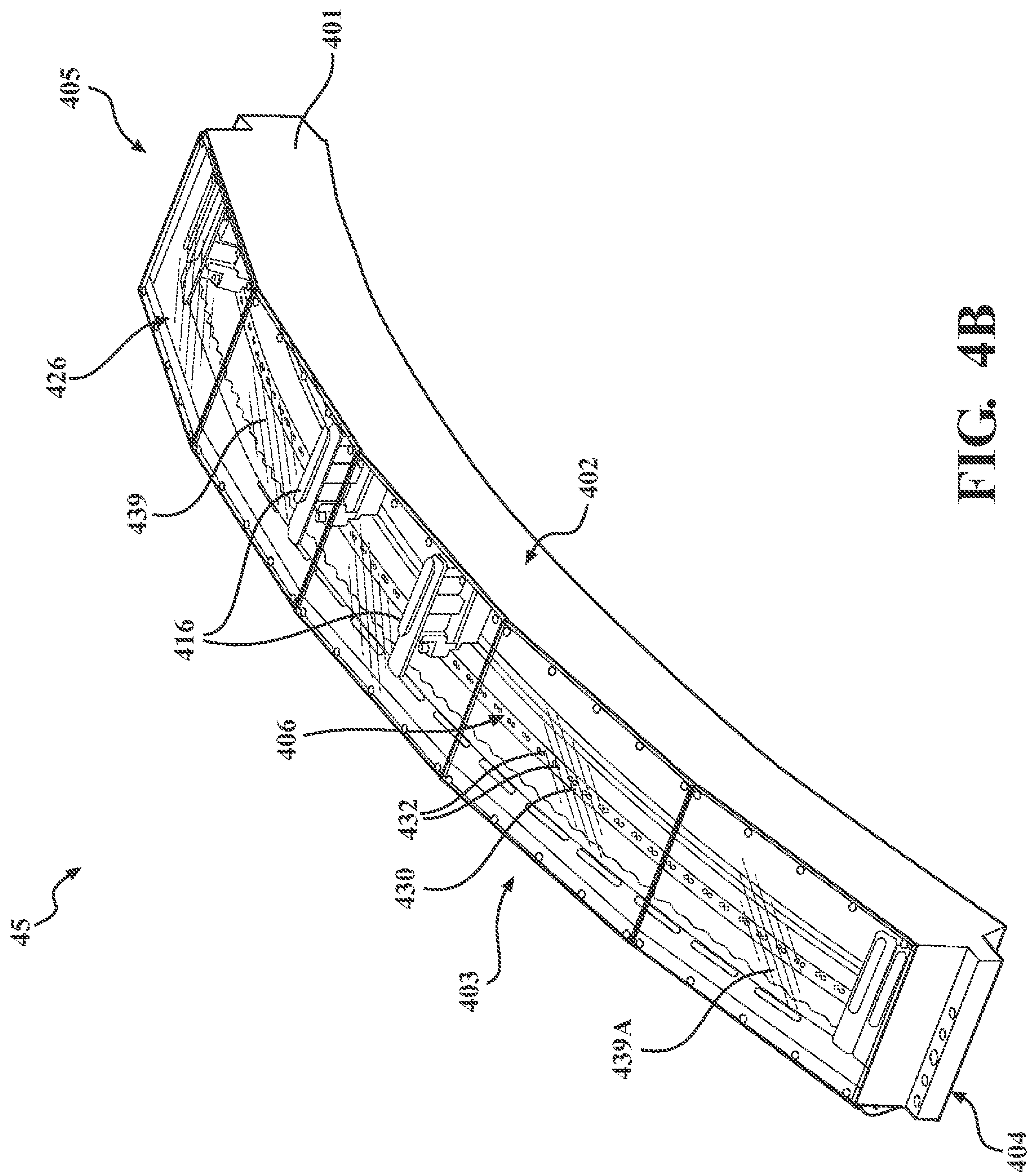

FIG. 4B

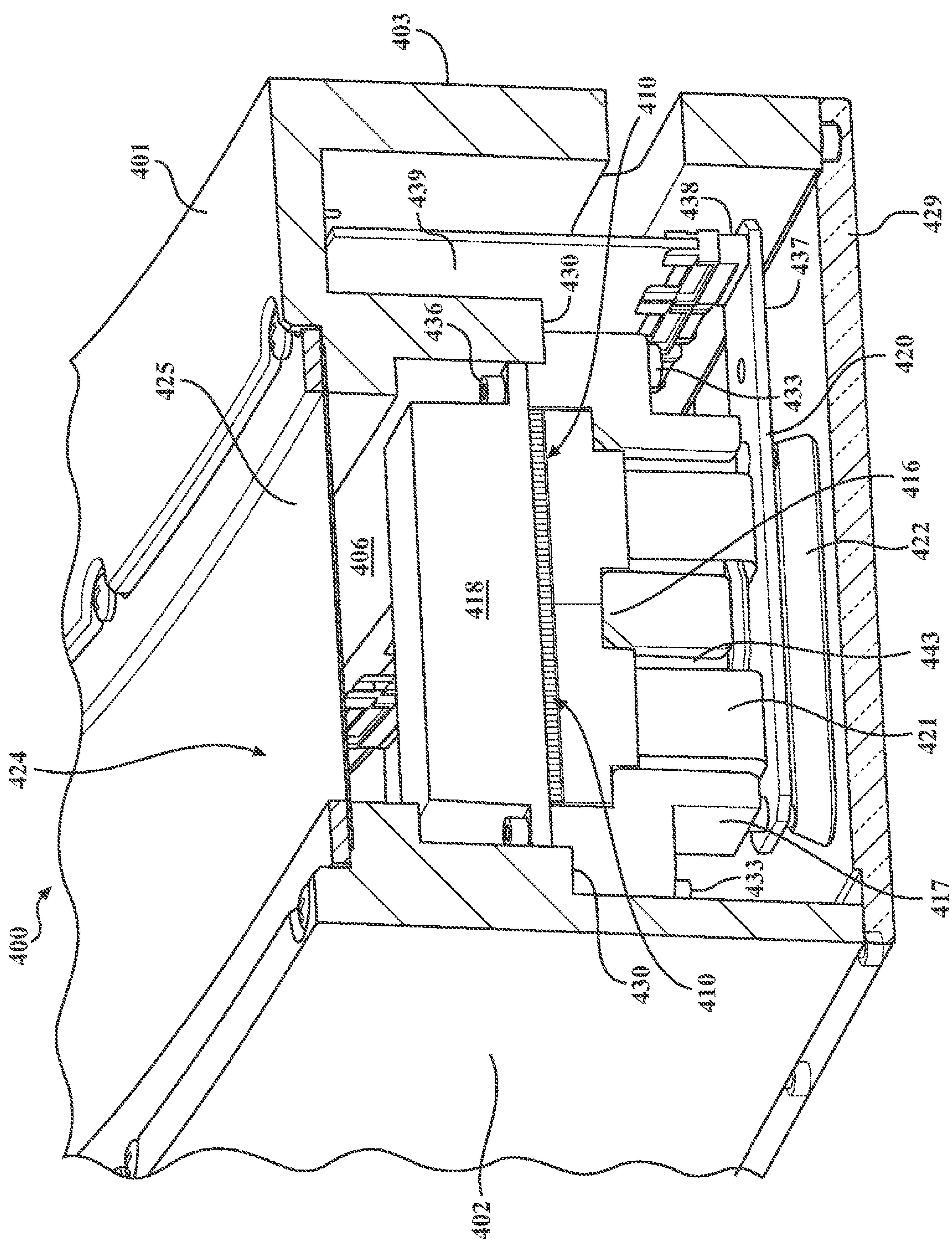

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FIG. 4C

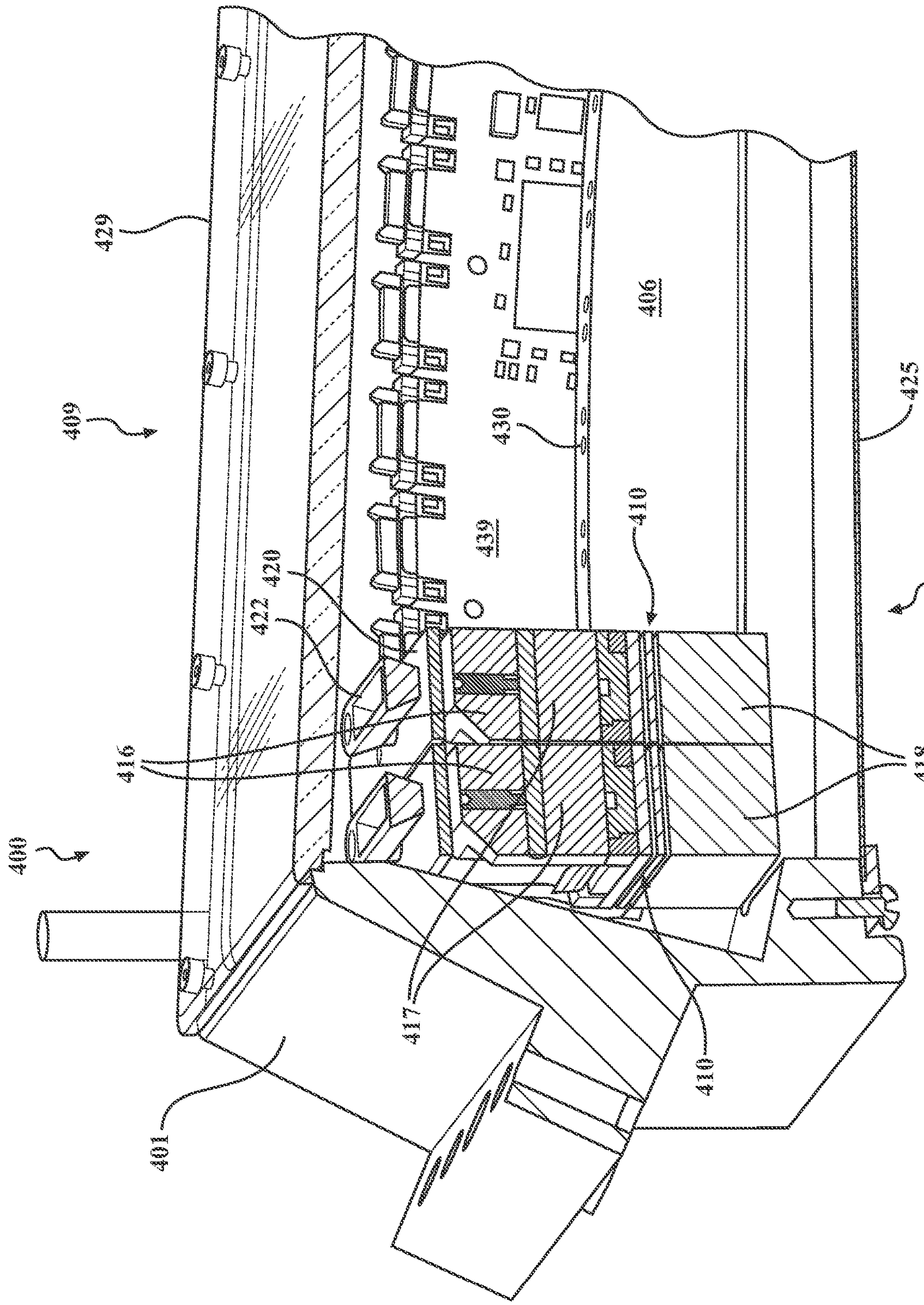

FIG. 4D

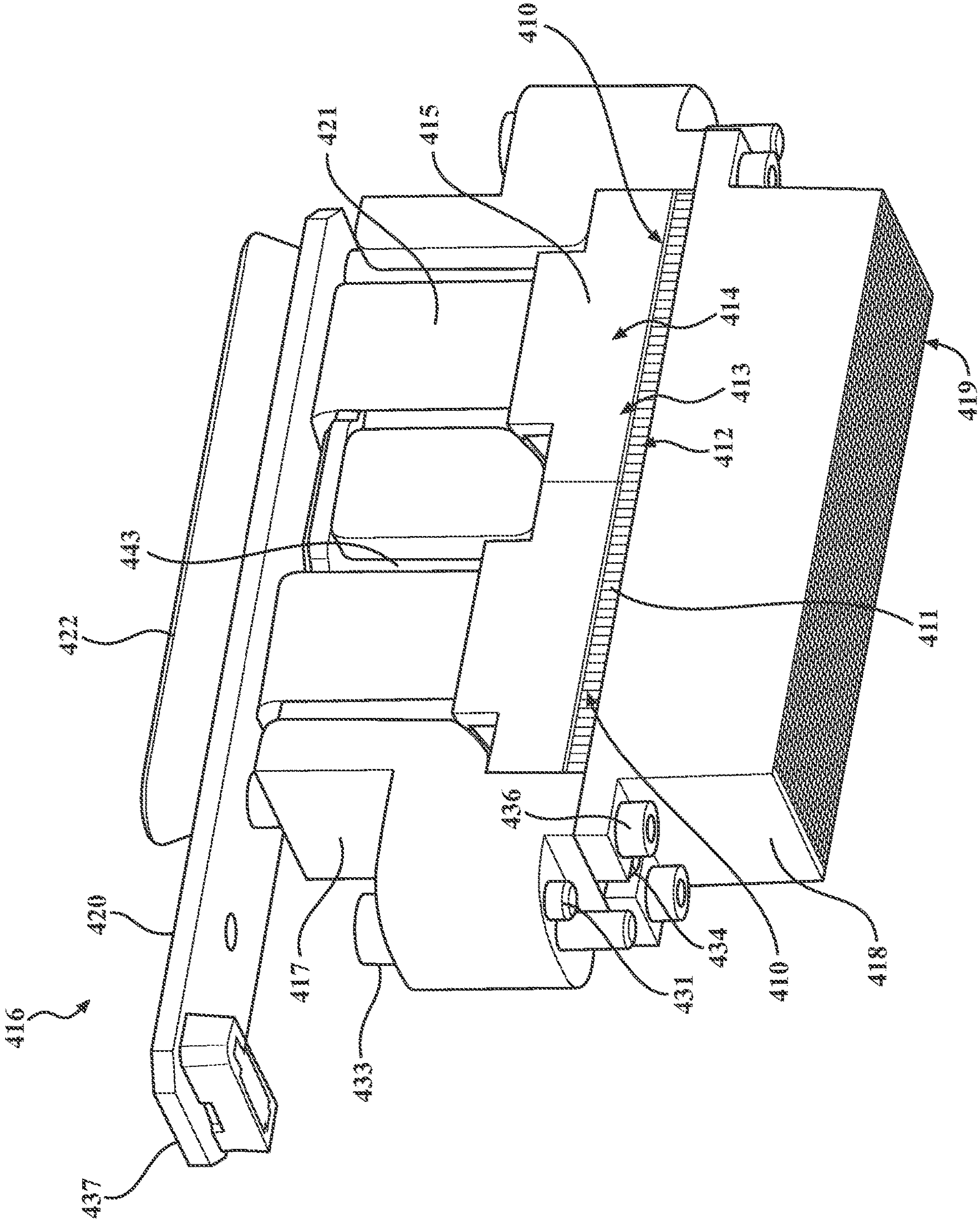
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FIG. 4E

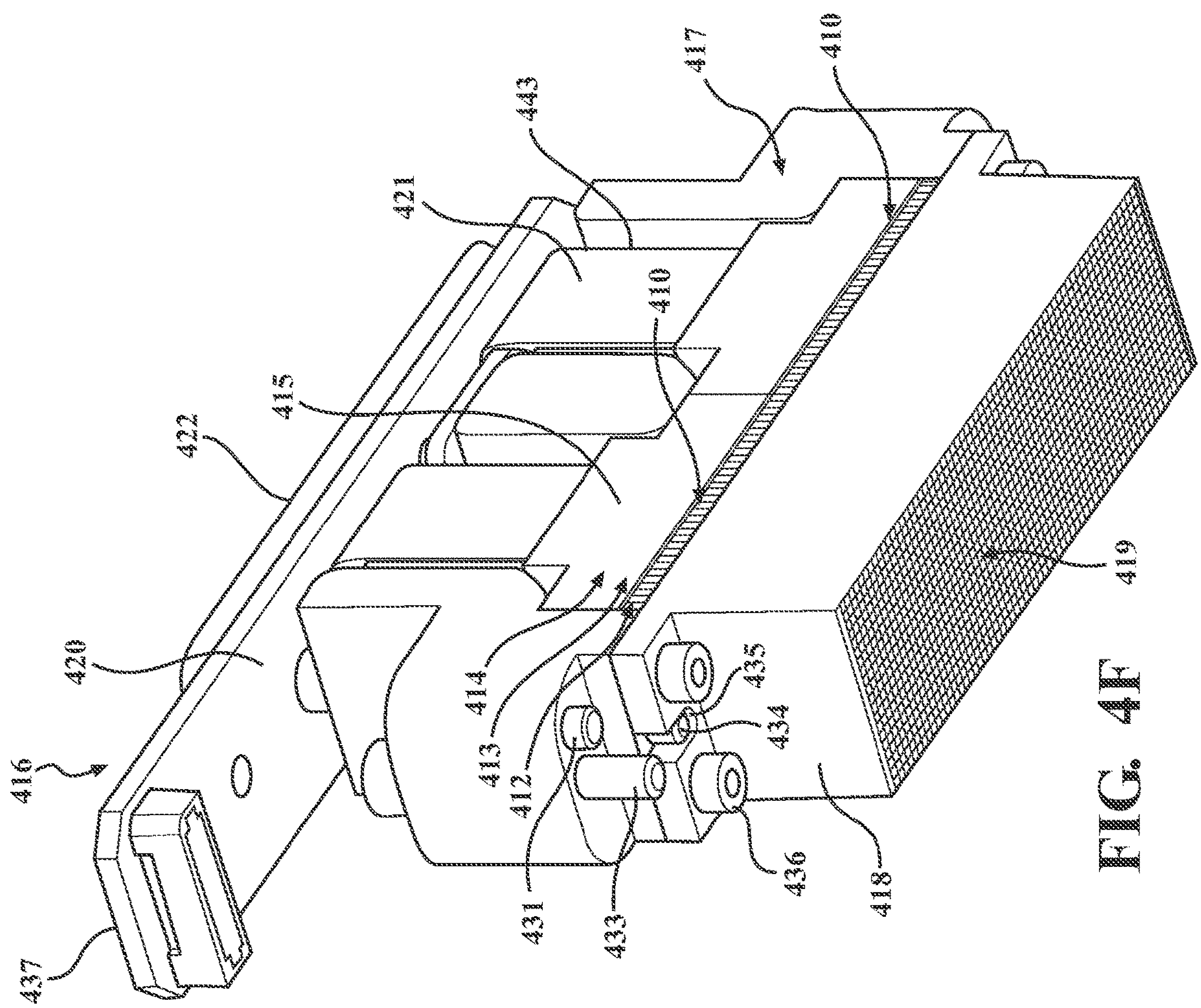
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FIG. 4F

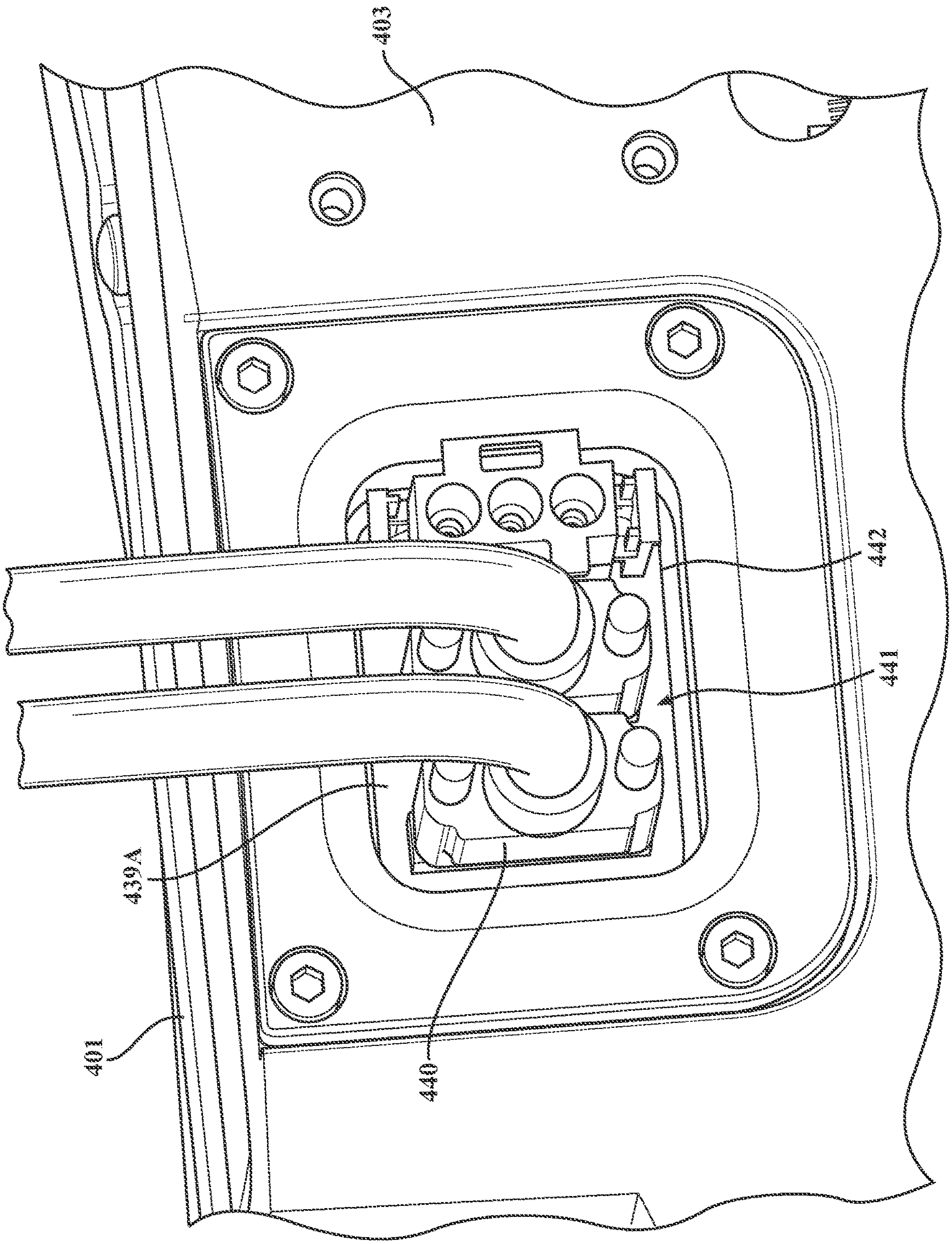

FIG. 4G

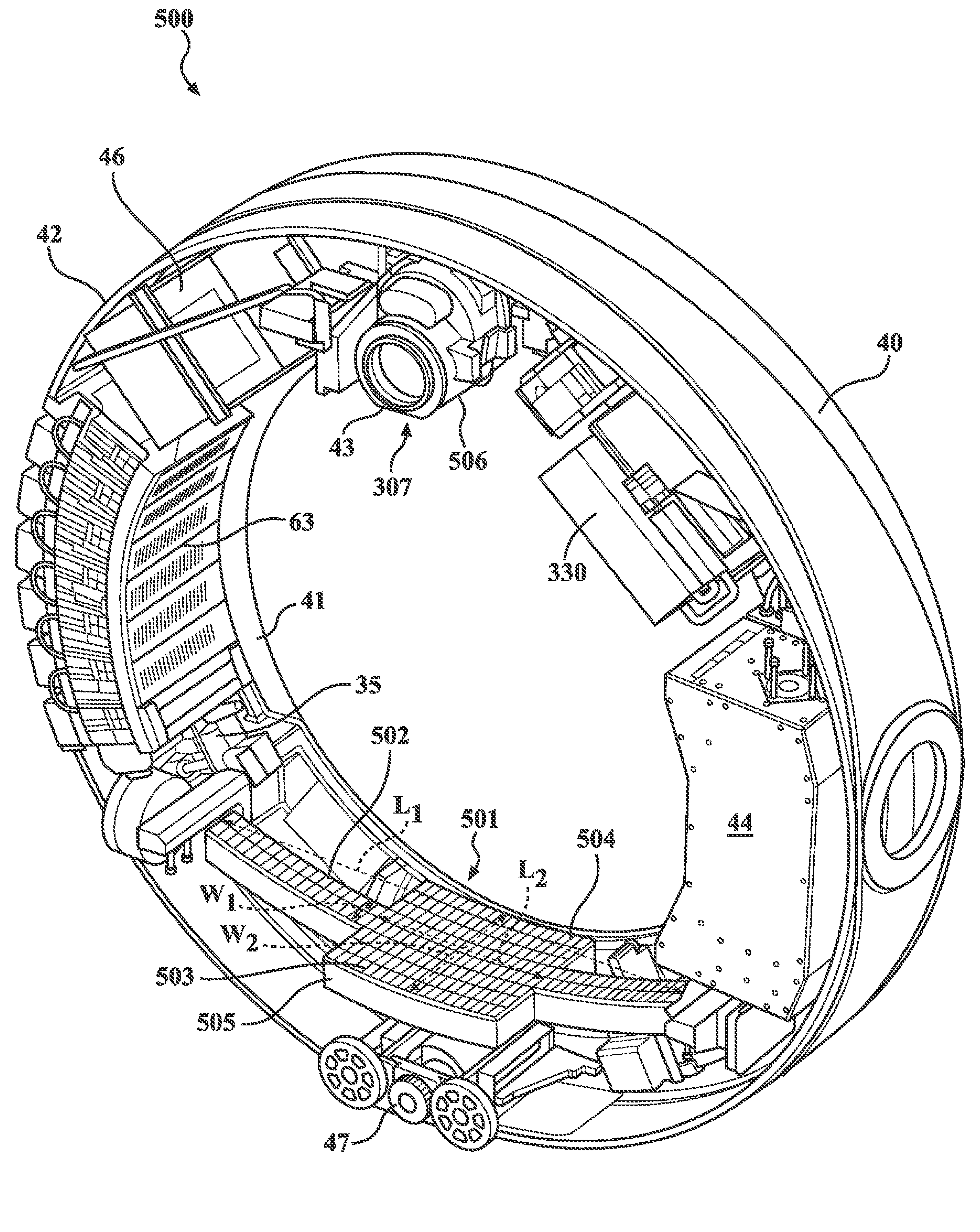
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FIG. 5A

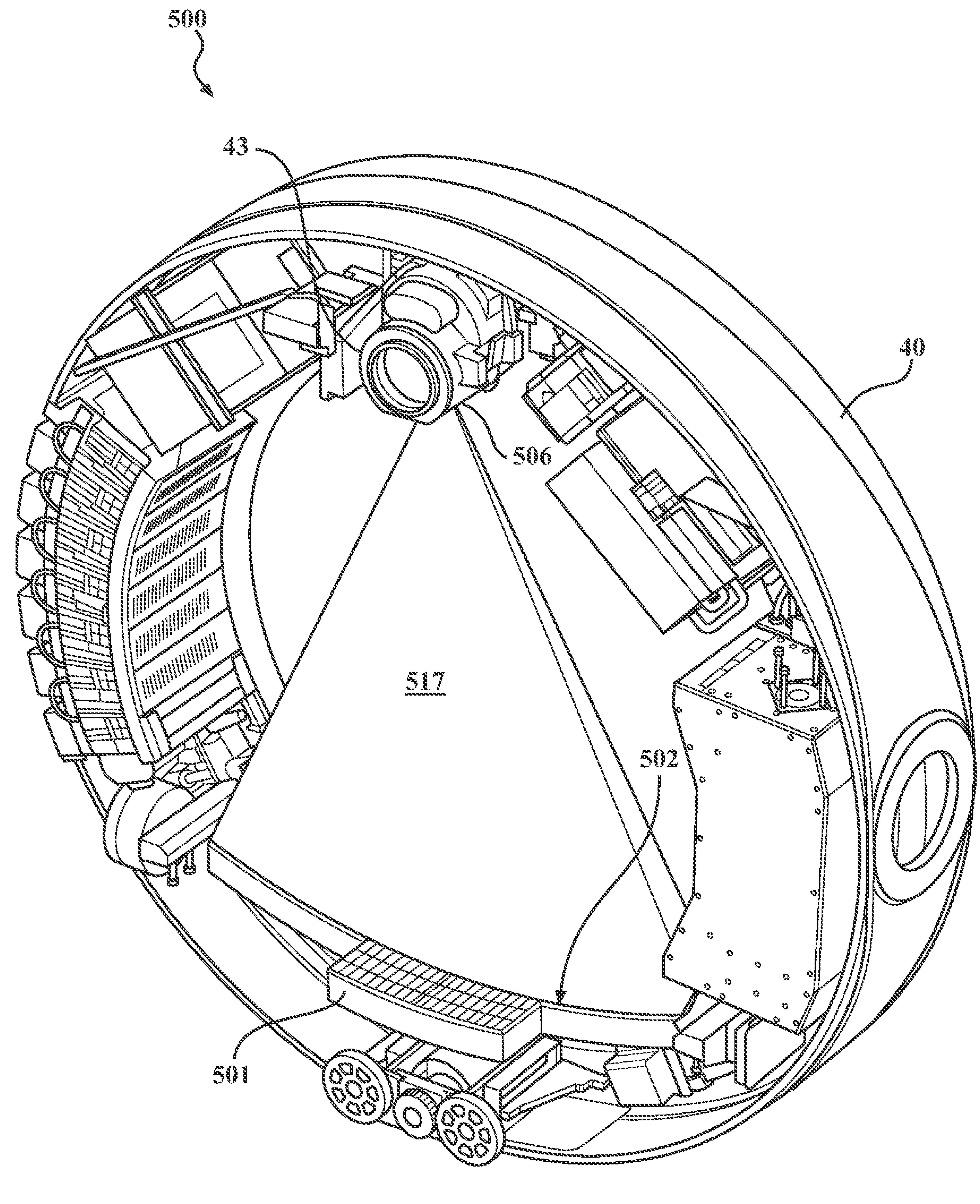

FIG. 5B

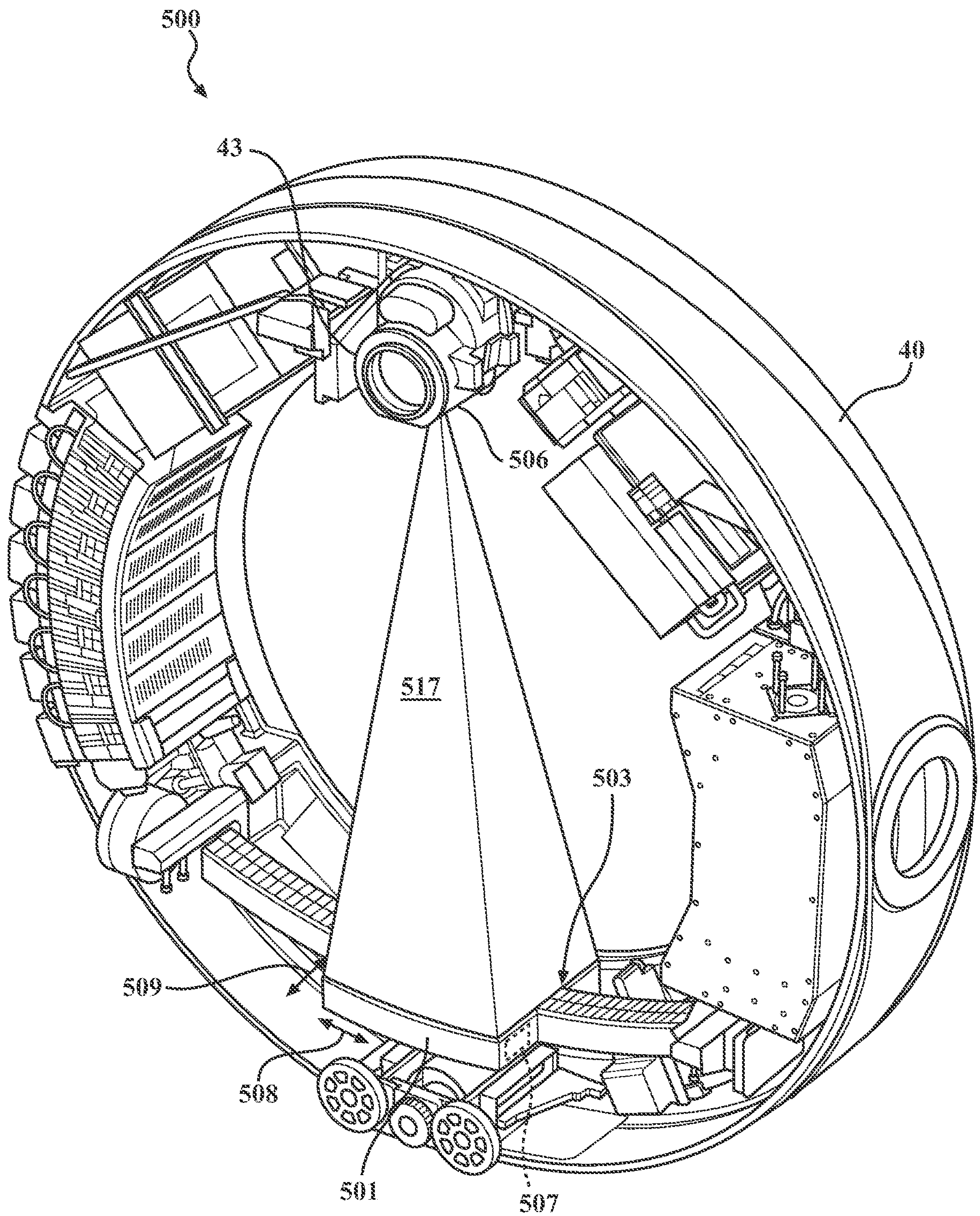

FIG. 5C

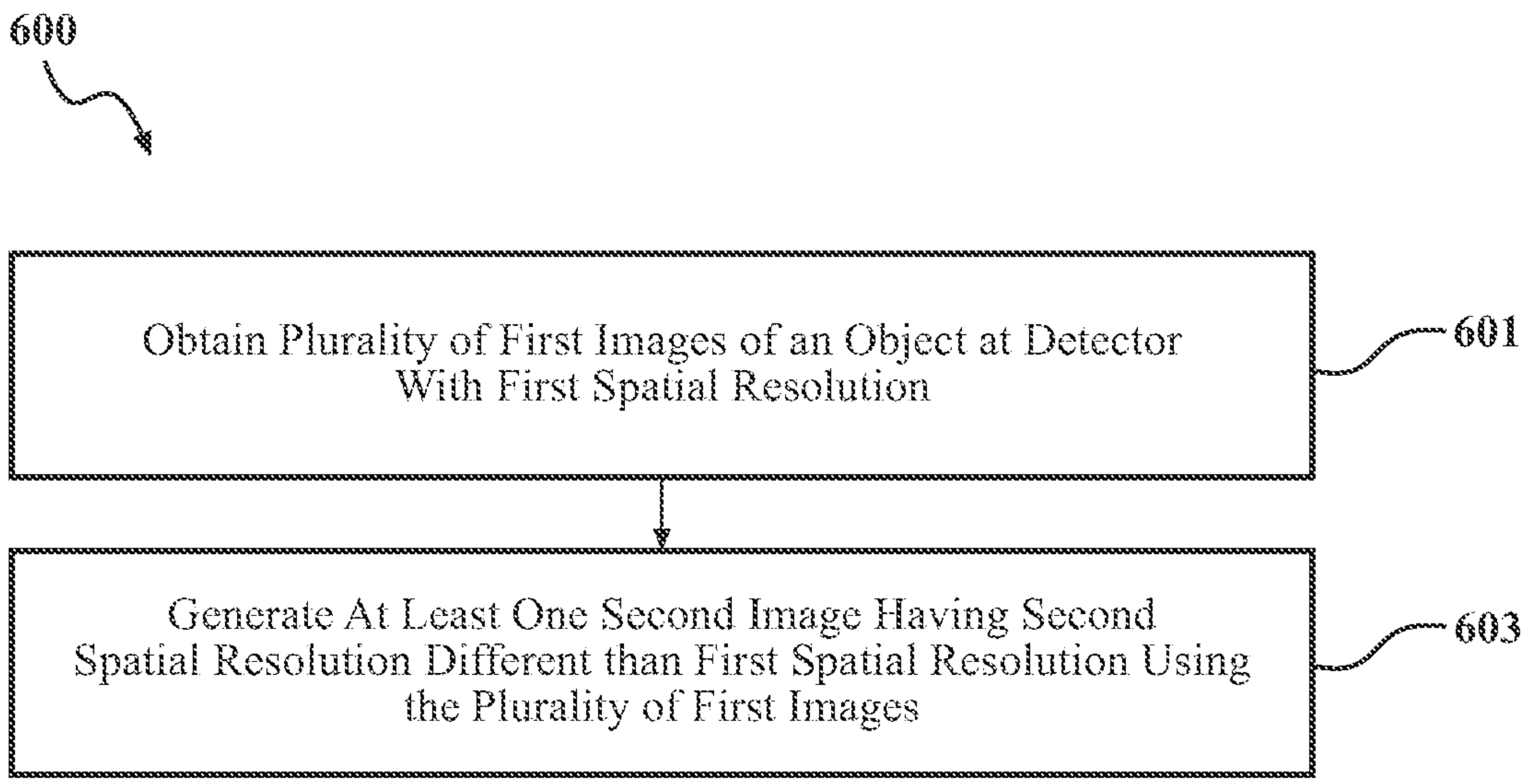
600
Obtain Plurality of First Images of an Object at Detector With First Spatial Resolution
601
Generate At Least One Second Image Having Second Spatial Resolution Different than First Spatial Resolution Using the Plurality of First Images
603

FIG. 6

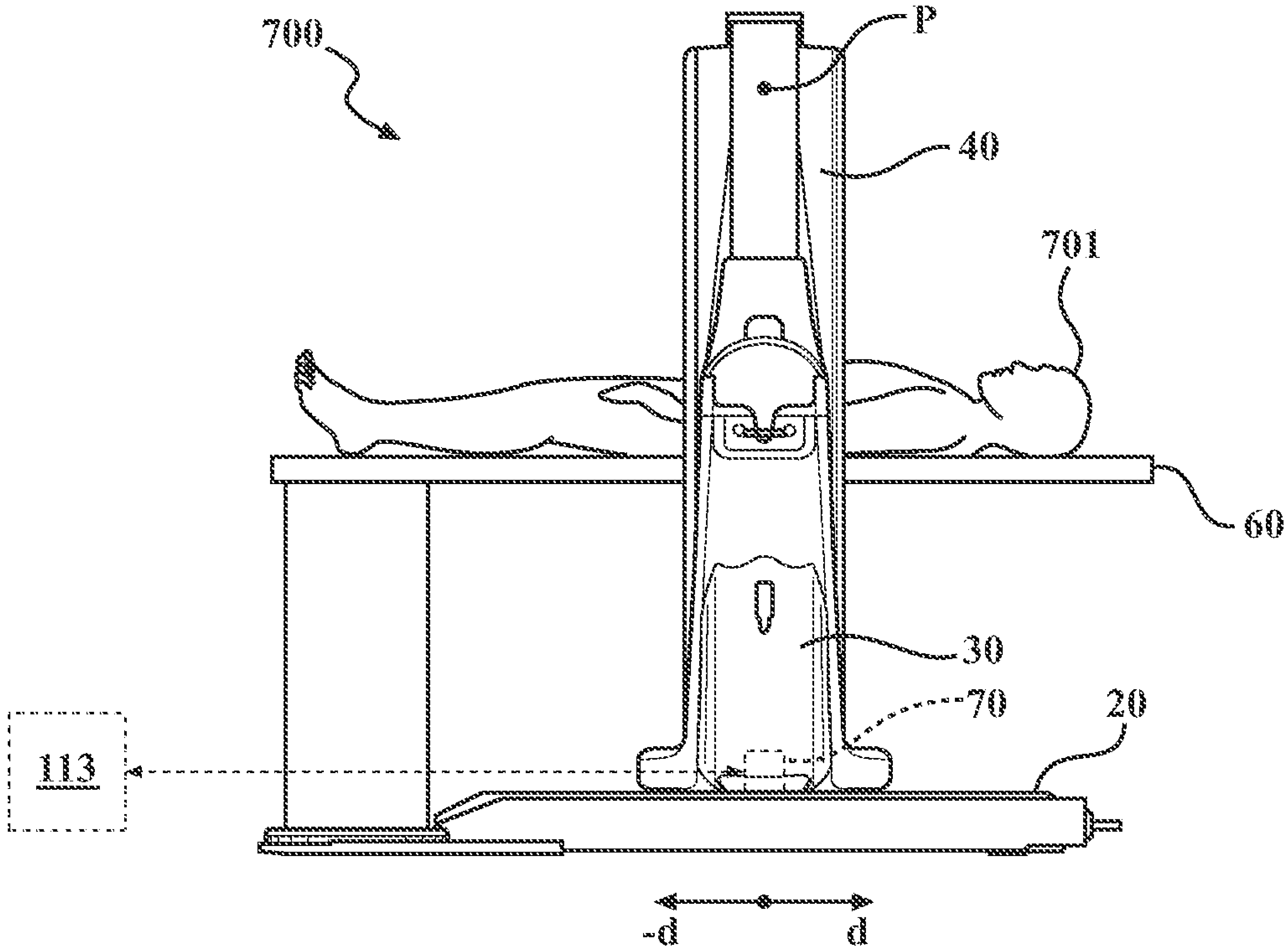

FIG. 7A

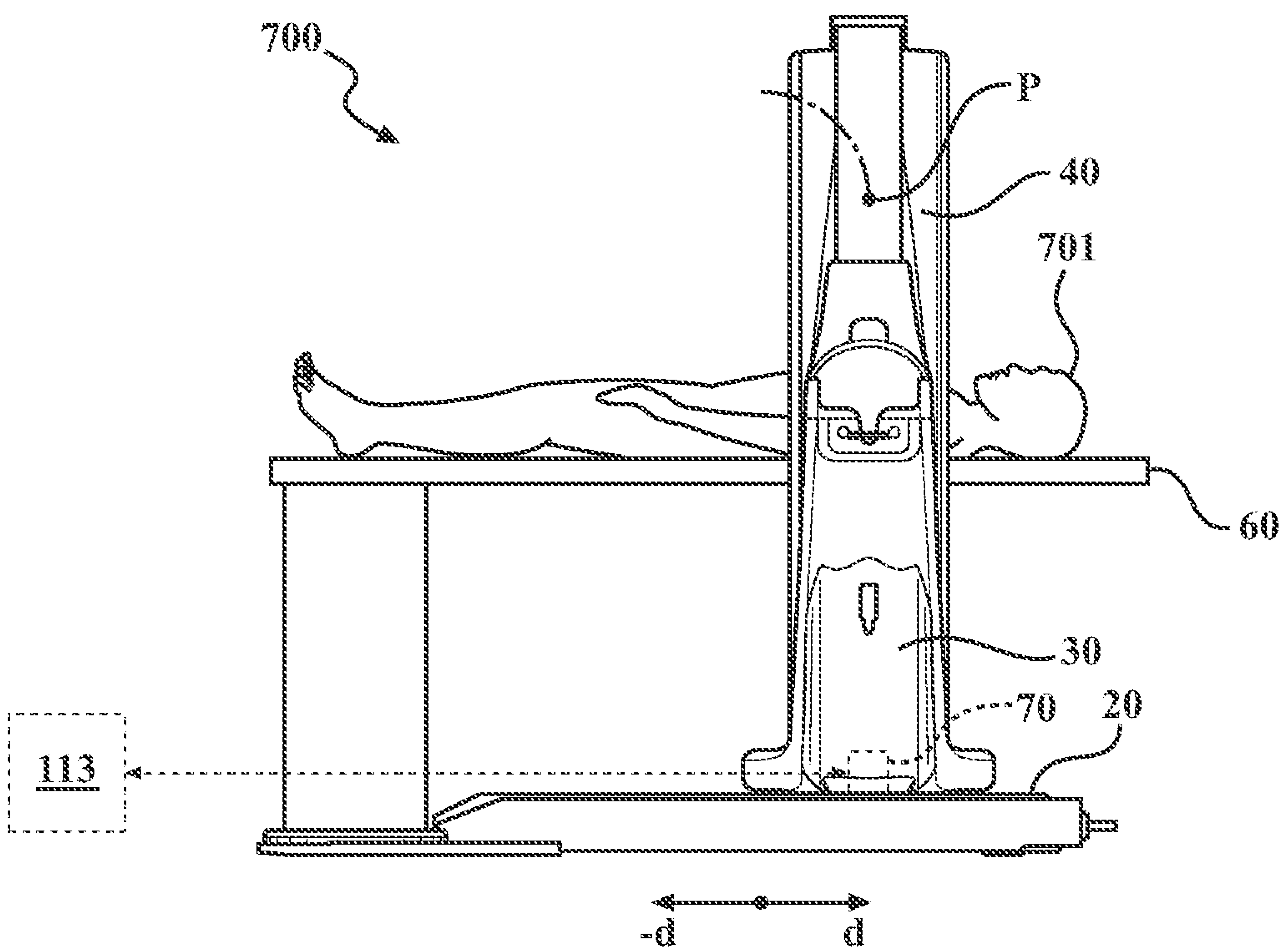

FIG. 7B

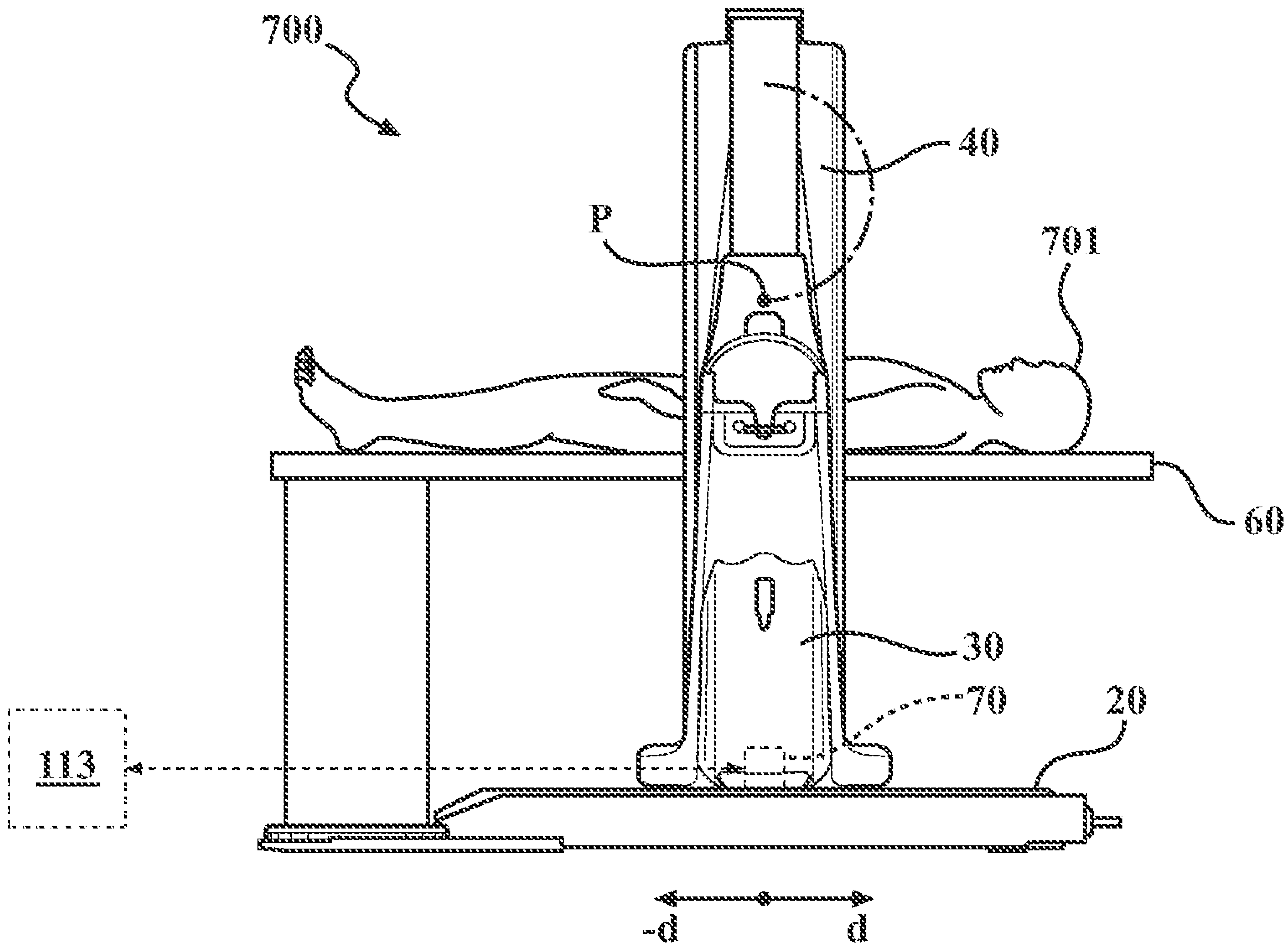

FIG. 7C

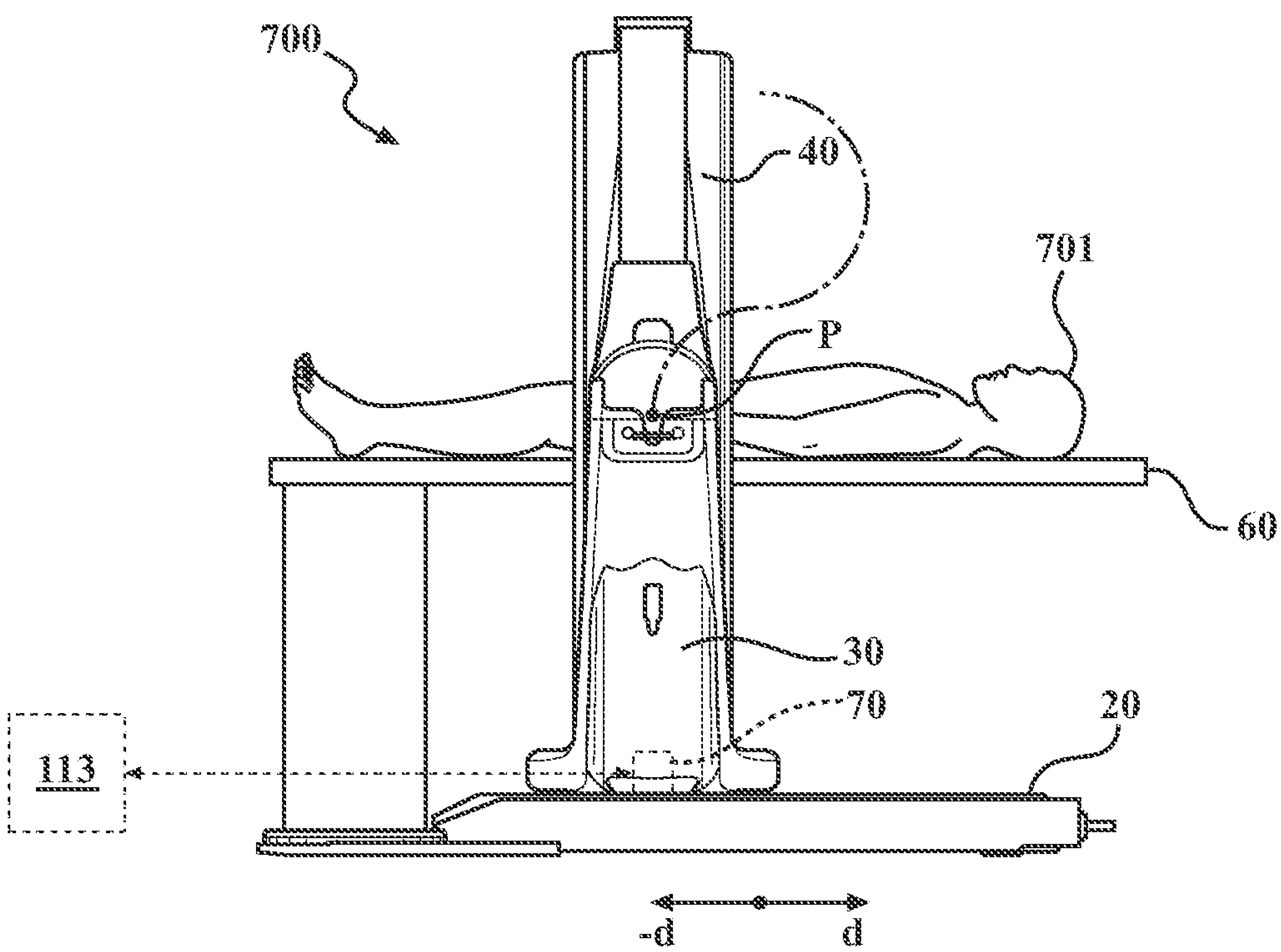

FIG. 7D

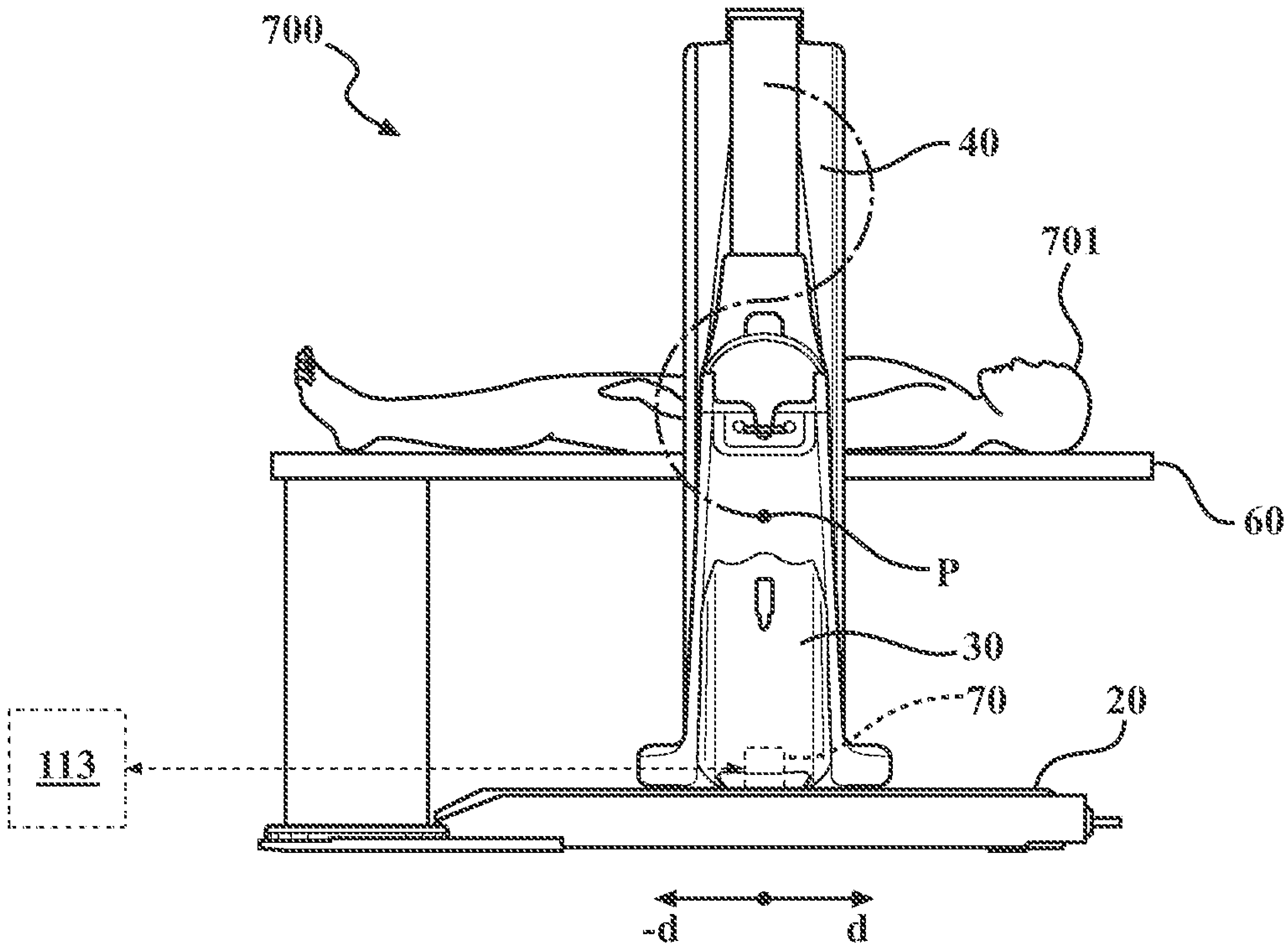

FIG. 7E

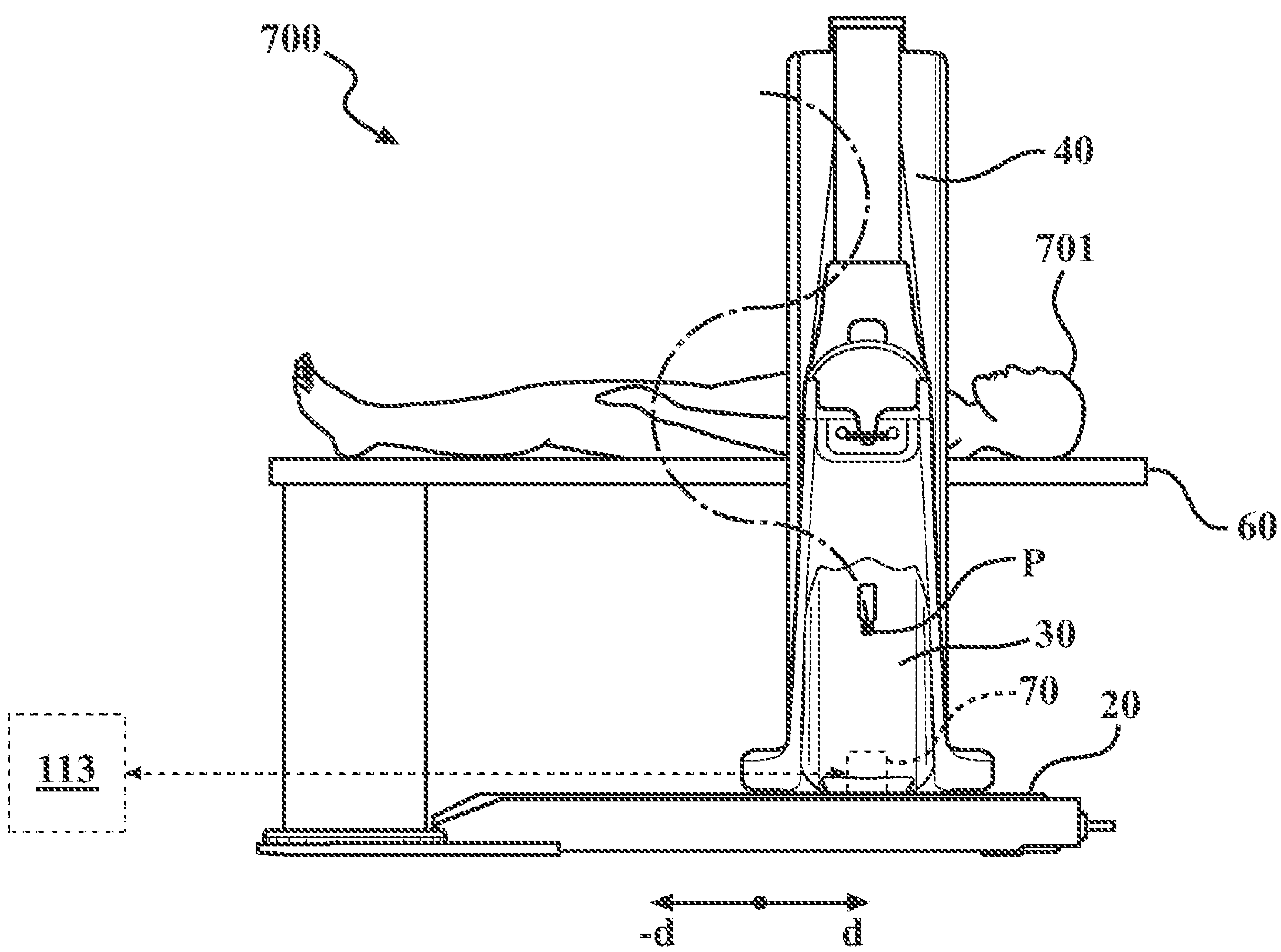

FIG. 7F

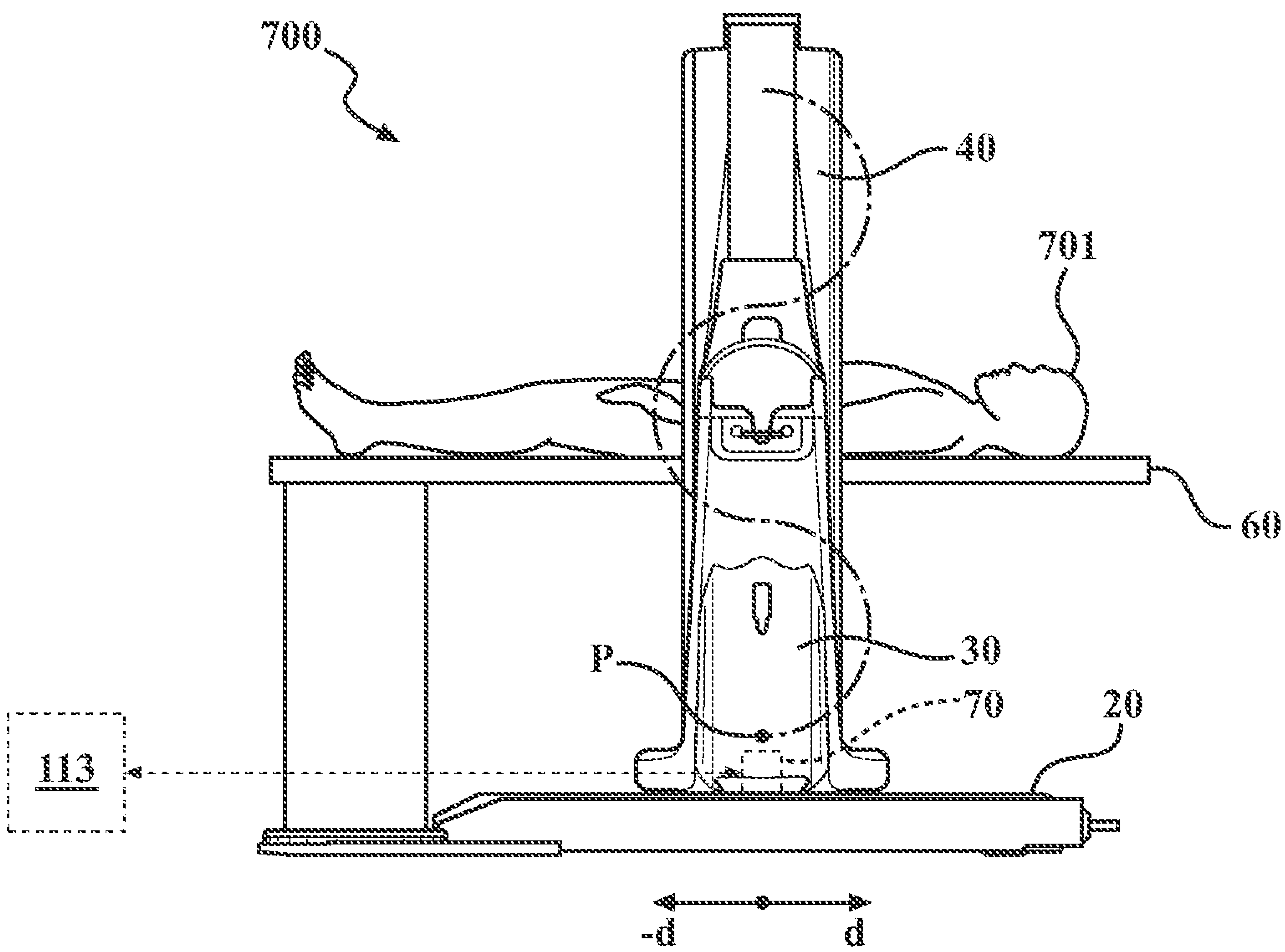

FIG. 7G

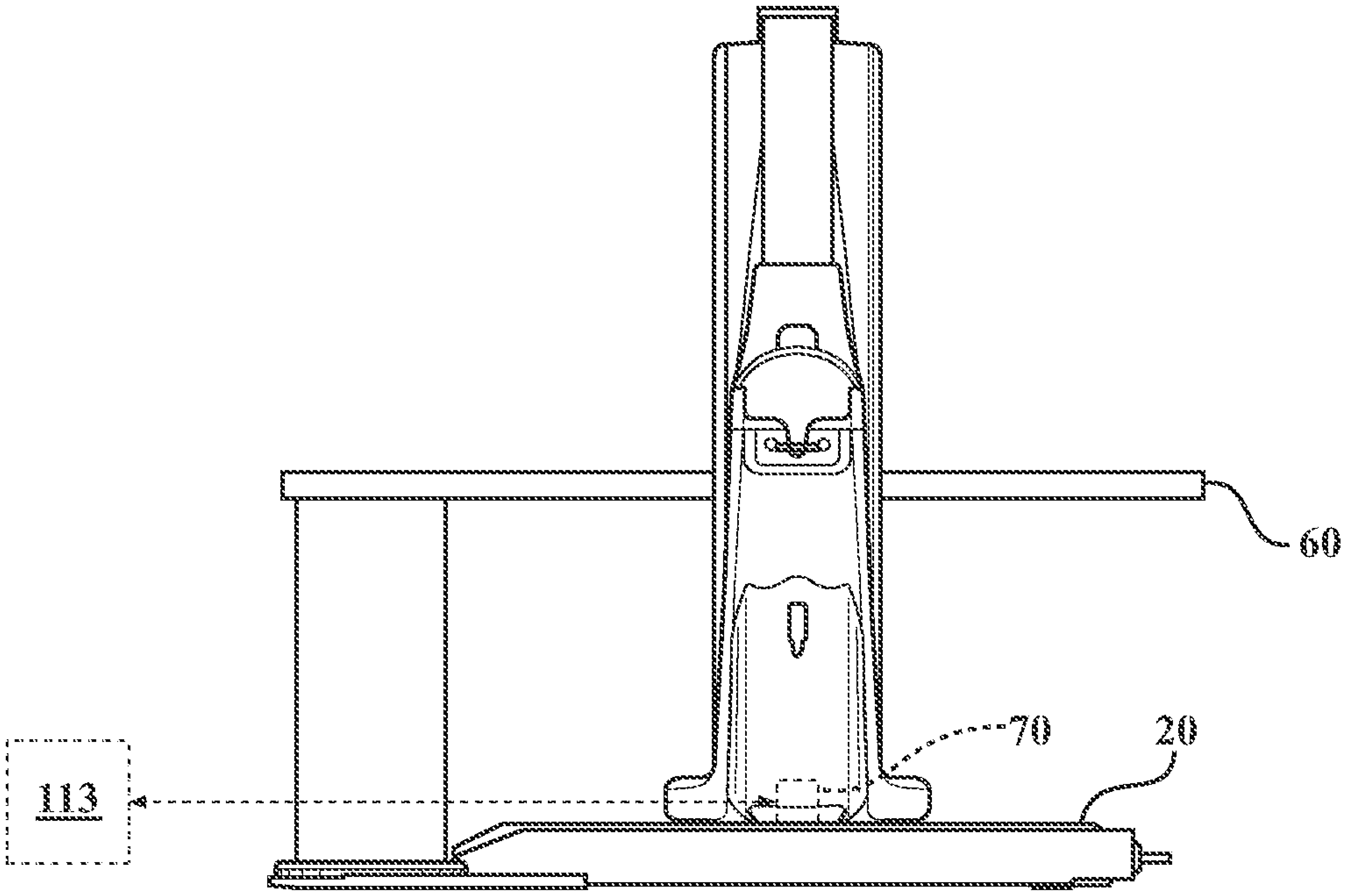

FIG. 8A

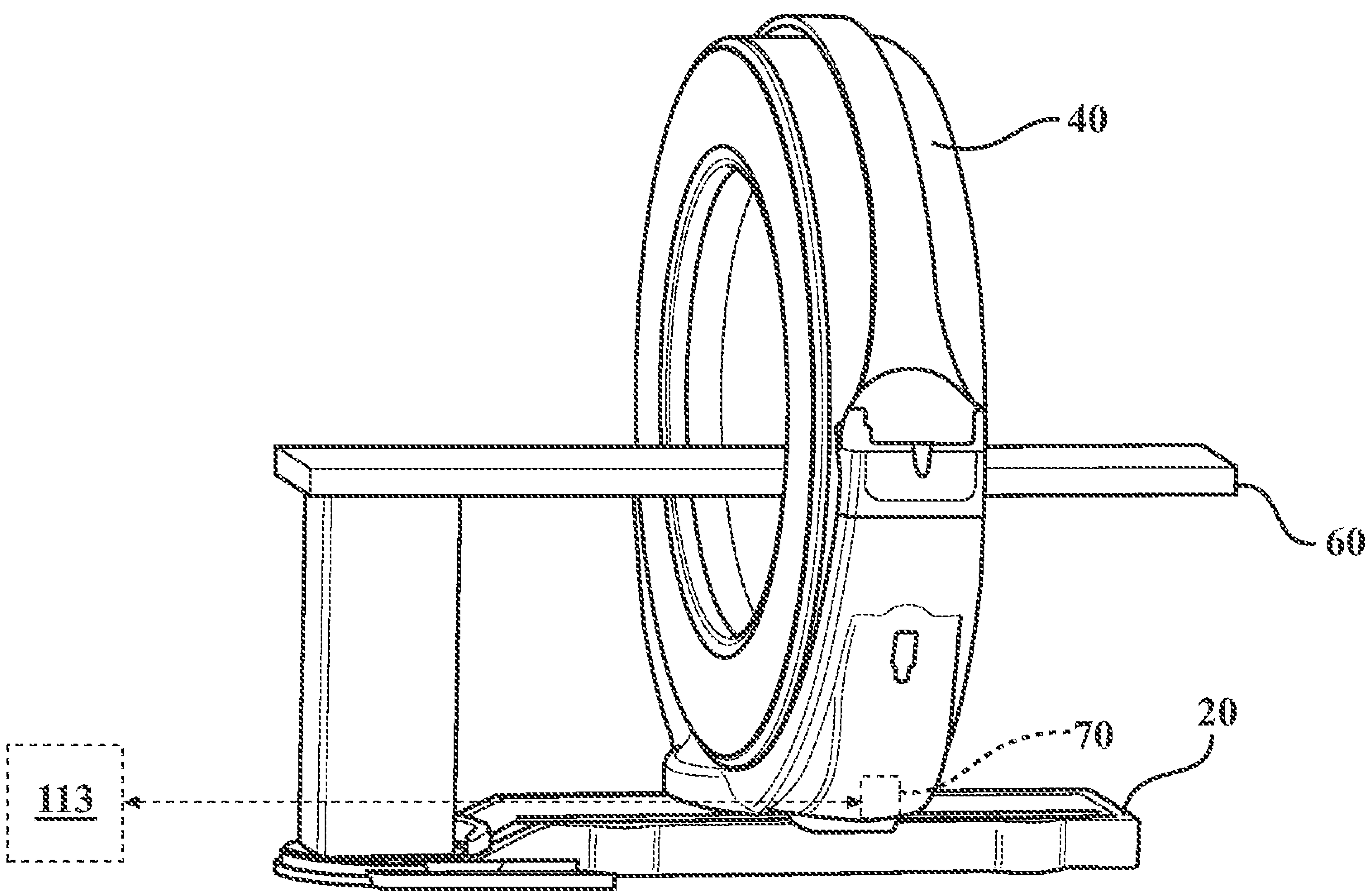

FIG. 8B

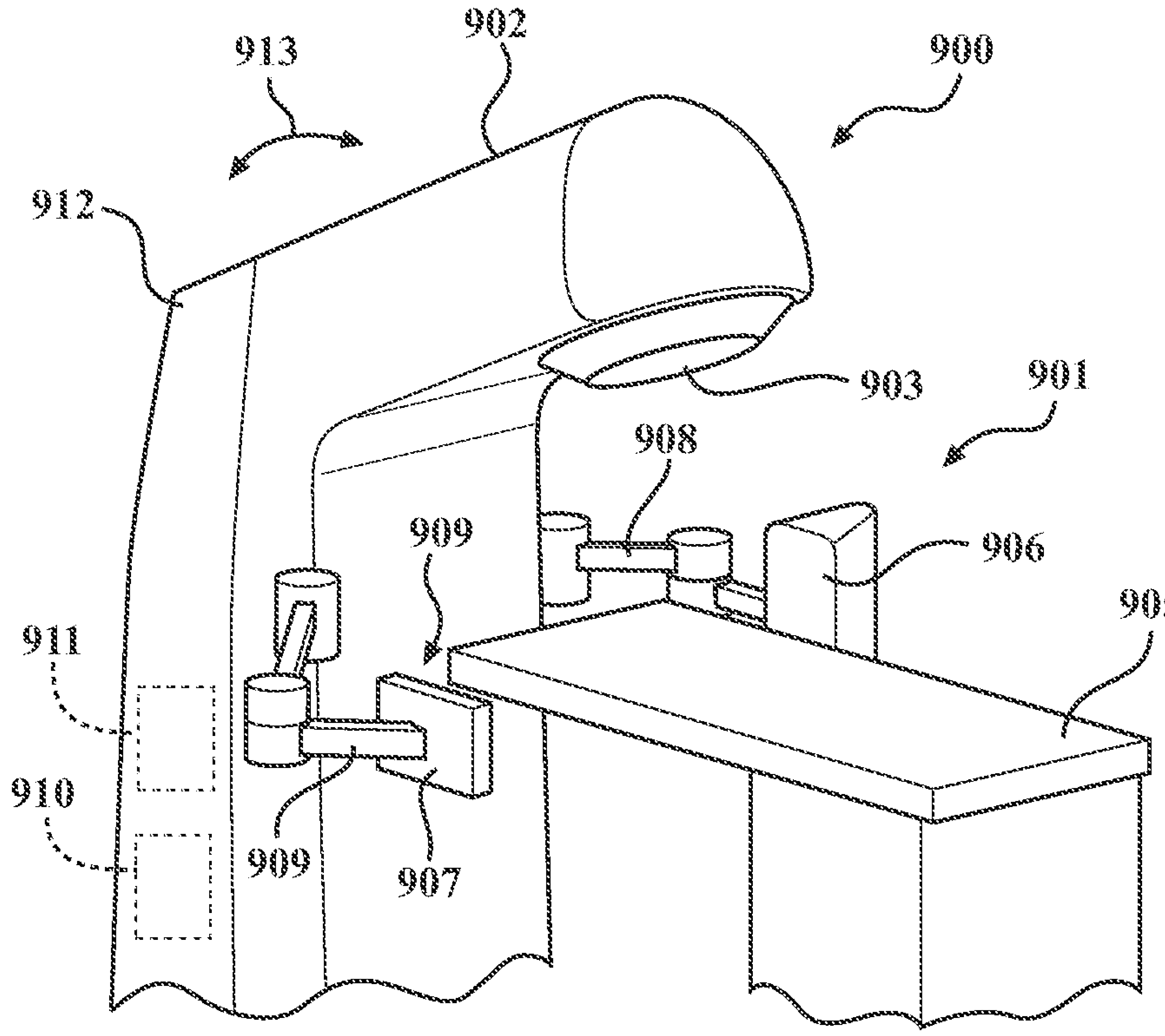
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FIG. 9

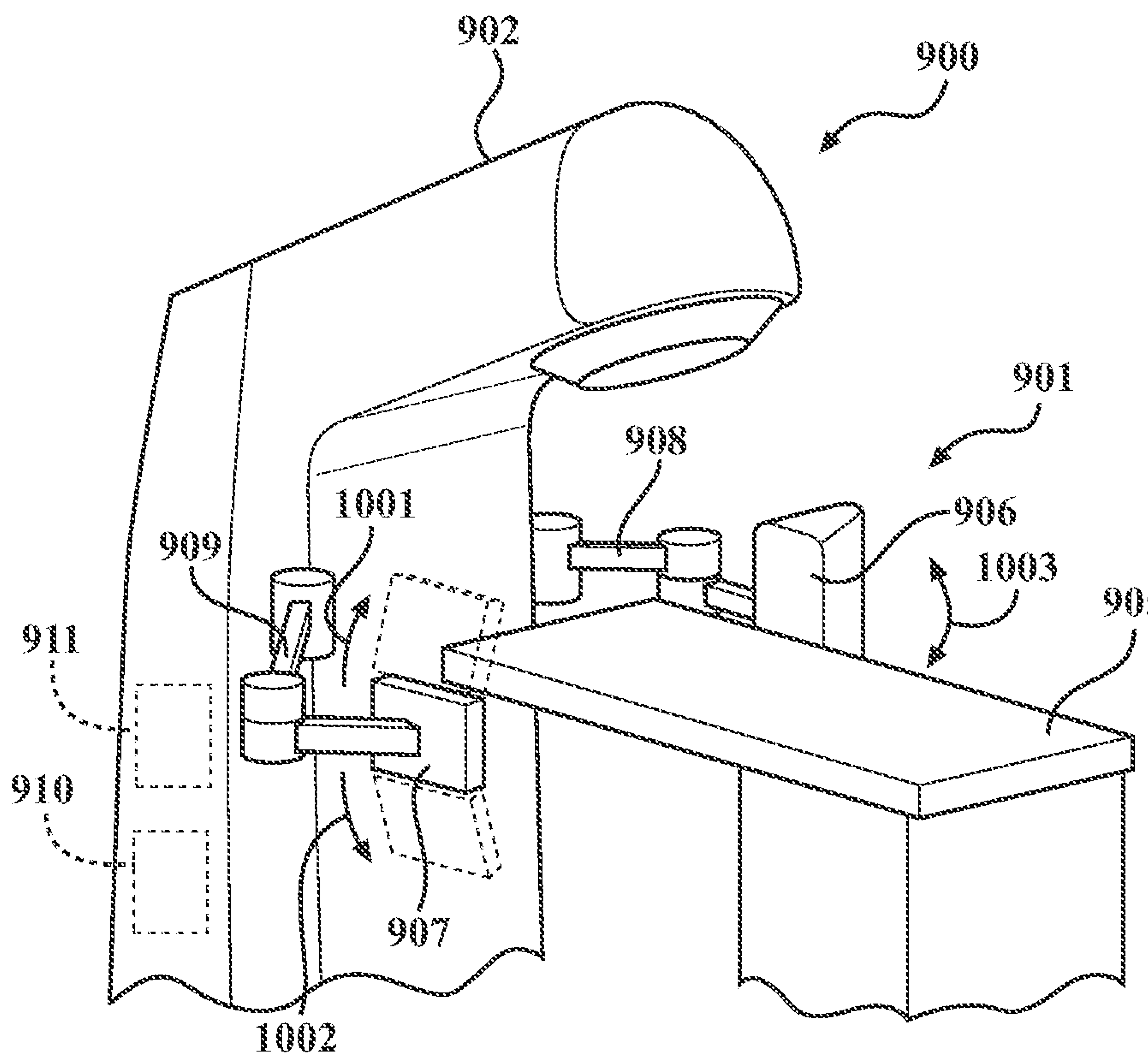
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FIG. 10

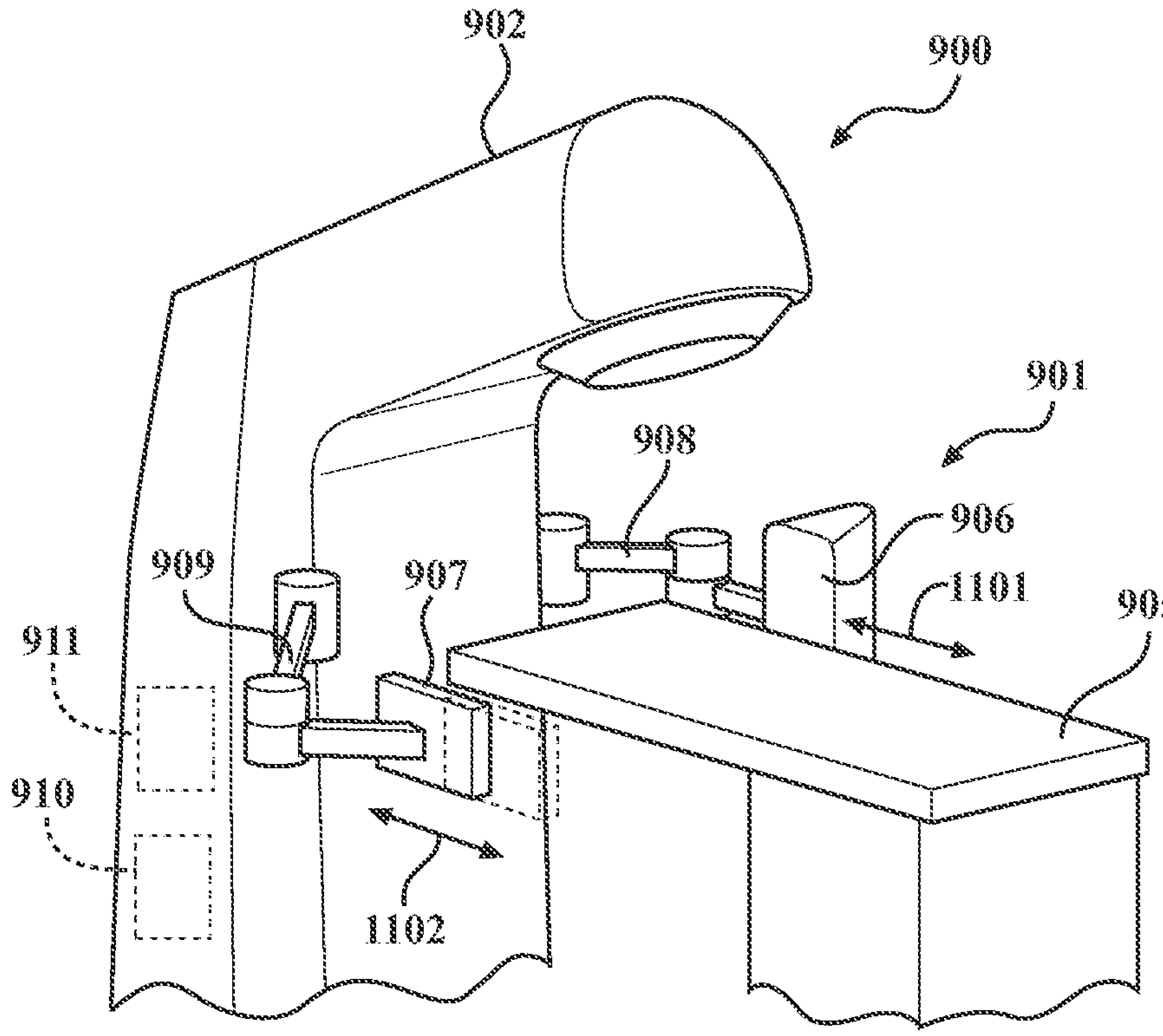
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FIG. 11

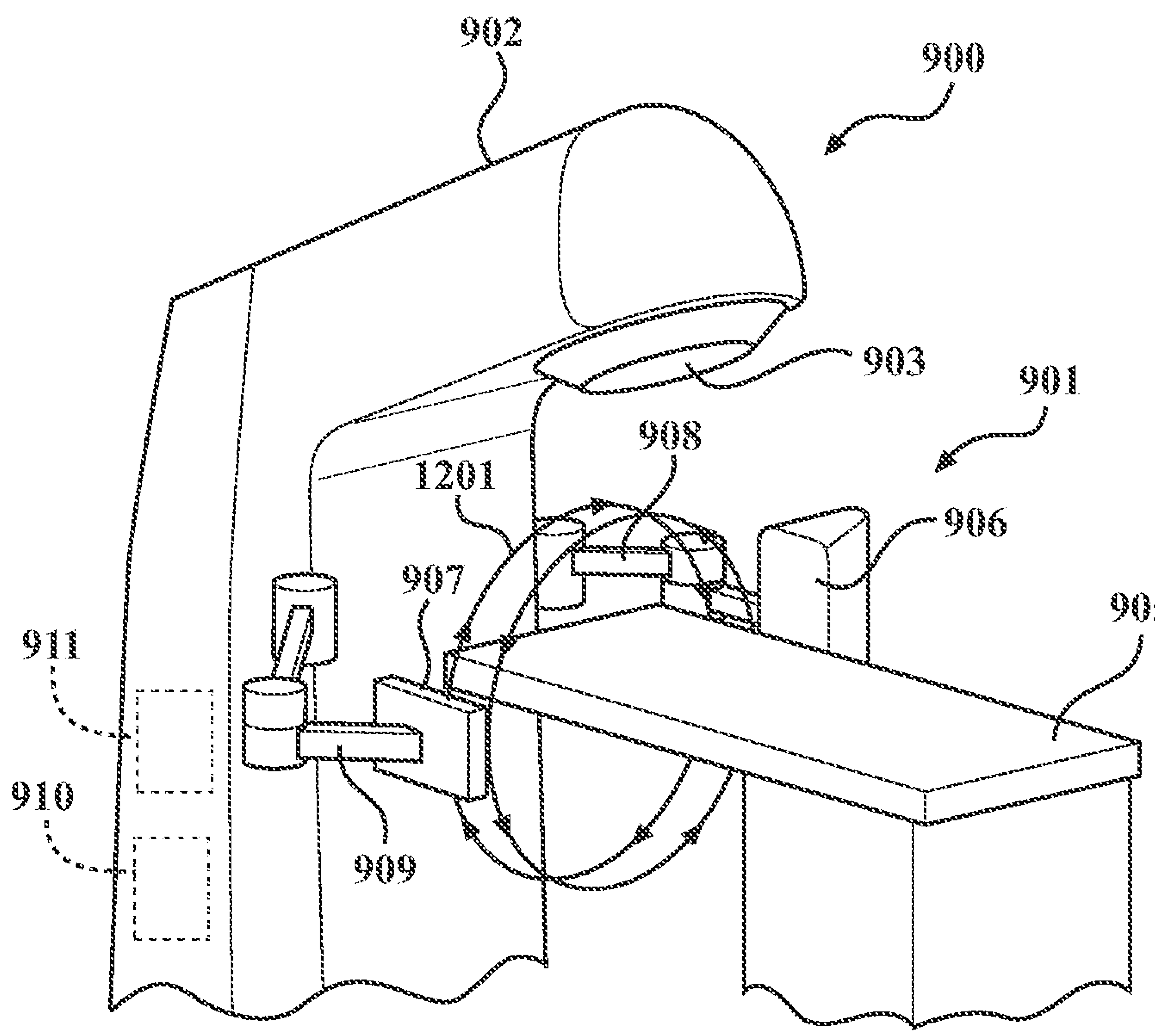
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FIG. 12

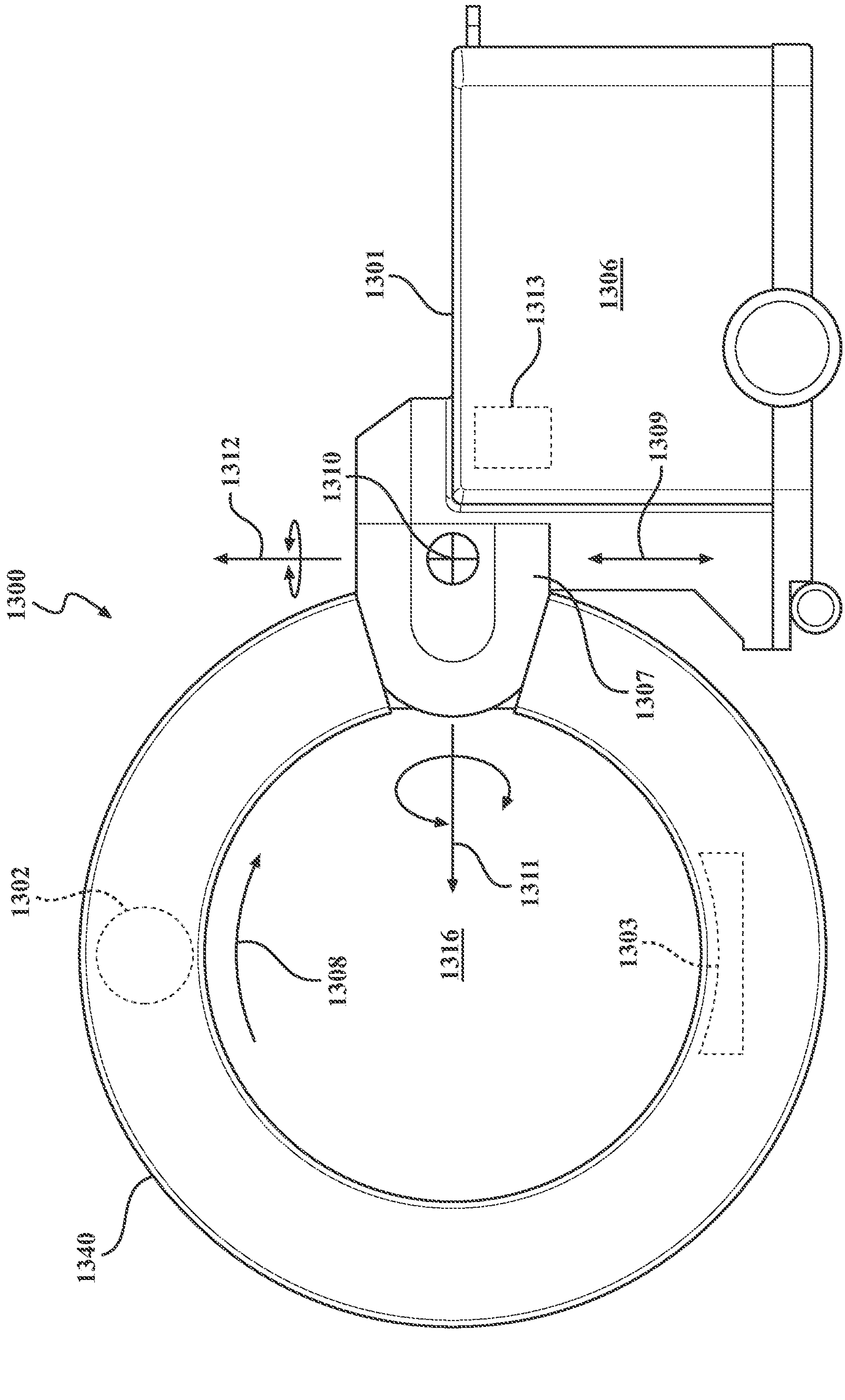
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FIG. 13

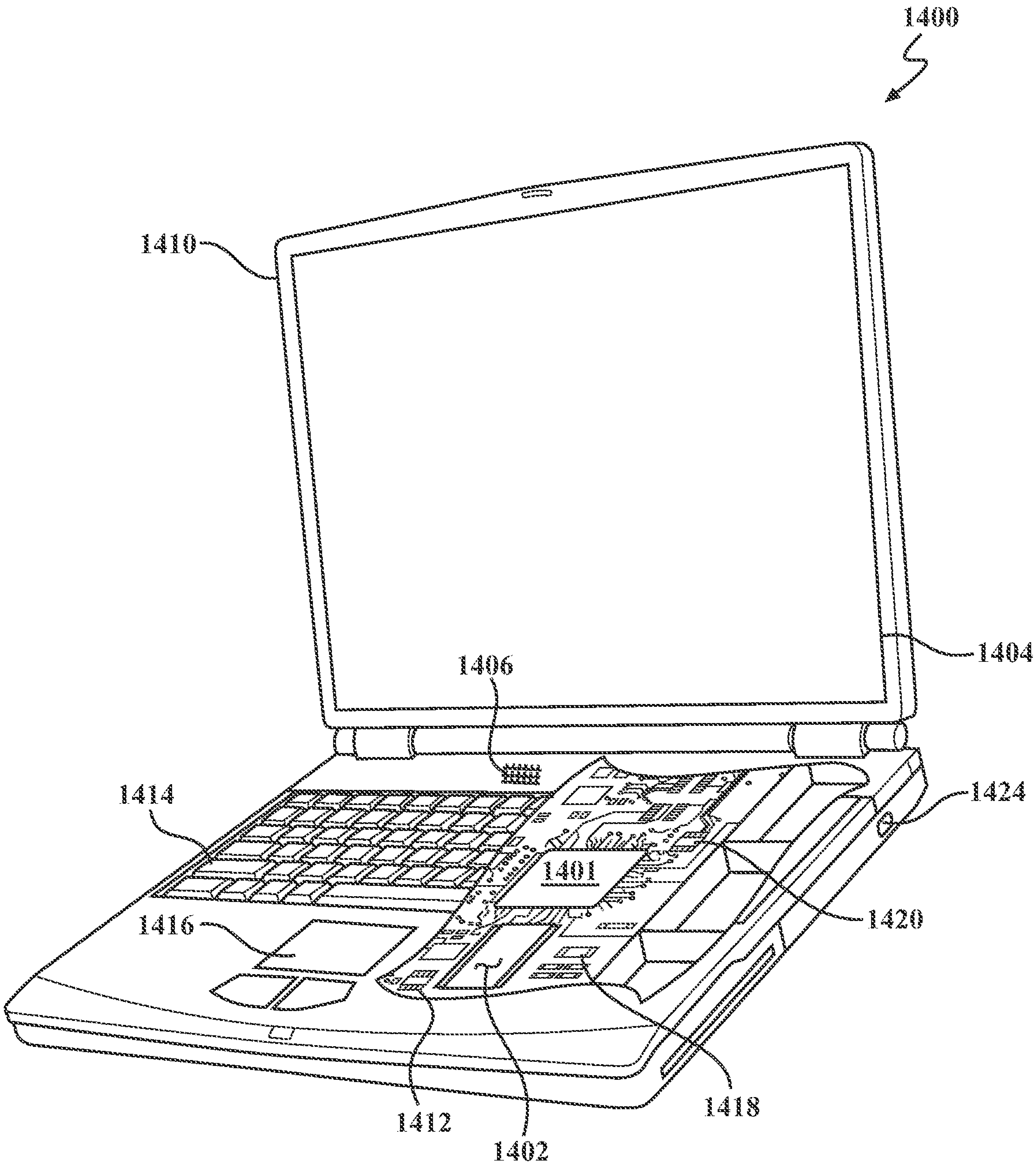

FIG. 14

MEDICAL X-RAY IMAGING SYSTEMS AND METHODS

RELATED APPLICATIONS

This application is a Continuation of U.S. patent application Ser. No. 16/354,047, filed on Mar. 14, 2019, which claims priority to and the benefit of U.S. Provisional Patent Application No. 62/644,032, filed Mar. 16, 2018, the disclosures of each of which are hereby incorporated by reference in their entirety.

BACKGROUND

Conventional medical imaging devices, such as x-ray computed tomography (CT) imaging devices, are limited in the types of imaging operations that may be performed.

SUMMARY

Embodiments include detector systems for an x-ray imaging device, x-ray imaging systems, external beam radiation treatment systems having an integrated x-ray imaging system, and methods therefor.

An embodiment includes a detector system for an x-ray imaging device that includes a detector chassis defining an interior housing, and a plurality of sub-assemblies mounted to the detector chassis within the interior housing, the plurality of sub-assemblies defining a detector surface, where each sub-assembly of the plurality of sub-assemblies includes a thermally-conductive support mounted to the detector chassis, a detector module including an array of x-ray sensitive detector elements mounted to a first surface of the support, an electronics board mounted to a second surface of the support that is opposite the first surface of the support, at least one electrical connector that connects the detector module to the electronics board, where the electronics board provides power to the detector module and receives digital x-ray image data from the detector module via the at least one electrical connector.

Further embodiments include an x-ray imaging system that includes an O-shaped gantry comprising a housing and defining an imaging bore, an x-ray source located within the housing of the gantry, a detector system located within the housing of the gantry opposite the x-ray source, the detector system including a plurality of x-ray sensitive detector elements defining a contiguous detector area, the detector area having an elongated central portion having a length of at least about 1 meter and a pair of peripheral portions extending on either side of the central portion to define a panel region having a width of greater than 0.3 meters and a length that is less than 0.5 meters, and a drive system for rotating the x-ray source and the detector system around the imaging bore.

Further embodiments include an x-ray imaging system that includes an O-shaped gantry comprising a housing and defining an imaging bore, an x-ray source located within the housing of the gantry, a detector located within the housing of the gantry opposite the x-ray source, the detector including a two-dimensional array of pixels, the array having a length and a width dimension of greater than about 0.3 meters, an apparatus for moving the detector array relative to the x-ray source by a distance that is less than a spacing between adjacent pixels of the detector array, a drive system for rotating the x-ray source and the detector system around an object located in the imaging bore, a processing device, coupled to the detector, that is configured with processor-executable instructions to perform operations including receiving, from the detector, a plurality of first x-ray images of the object with a first spatial resolution that are obtained while the detector array is moved relative to the x-ray source, and generating at least one second image of the object using the plurality of first x-ray images, wherein the at least one second image is a super resolution (SR) image that has an improved spatial resolution and/or signal-to-noise (SNR) ratio compared to the first images.

Further embodiments include an x-ray imaging system that includes an O-shaped gantry comprising a housing and defining an imaging bore, a drive mechanism for moving the gantry with respect to a patient located within the imaging bore of the gantry, an x-ray source located within the housing of the gantry, a detector located within the housing of the gantry opposite the x-ray source, the detector including a two-dimensional array of pixels, the array having a length and a width dimension of greater than about 0.3 meters, a rotation drive system configured to rotate the x-ray source and the detector around the patient located in the imaging bore, and a control system, coupled to the drive mechanism, and including a processor that is configured with processor-executable instructions to perform operations including controlling the drive mechanism to move the x-ray source and detector system relative to the patient located within the bore of the gantry as the x-ray source and detector system rotate around the patient so that the x-ray source and detector system follow a sinusoidal scan trajectory around the patient.

Further embodiments include an x-ray imaging system that includes an O-shaped gantry including a housing and defining an imaging bore, a drive mechanism configured to translate the gantry with respect to a patient located within the imaging bore, an x-ray source located within the housing of the gantry, a detector located within the housing of the gantry opposite the x-ray source, the detector including a two-dimensional array of pixels, the array having a length and a width dimension of greater than about 0.3 meters, a rotation drive system configured to rotate the x-ray source and the detector around the patient located in the imaging bore, and a control system, coupled to the drive mechanism and to the rotation drive system, and including a processor that is configured with processor-executable instructions to perform operations including controlling the rotation drive system to rotate the x-ray source and the detector around a patient located in the bore of the gantry in a first rotational direction and then in a second rotational direction that is opposite the first direction; and controlling the drive mechanism to translate the gantry in a first translation direction along the length of the patient as the x-ray source and detector system rotate around the patient in the first rotational direction and in the second rotational direction so that the x-ray source and detector follow a reverse helical scan trajectory around the patient.

Further embodiments include an external beam radiation treatment system that includes a rotatable gantry having a head that emits a radiation treatment beam, the rotatable gantry configured to rotate around a patient to direct the radiation treatment beam to a target location within the patient from different angles, an x-ray imaging system mounted to the rotatable gantry, the x-ray imaging system including an x-ray source, a detector opposite the x-ray source, a first actuator system for moving the x-ray source relative to the rotatable gantry, and a second actuator system for moving the detector relative to the rotatable gantry, and a control system, coupled to the first actuator system and to the second actuator system, and including a processor that is configured with processor-executable instructions to perform operations including controlling the first actuator system and the second actuator system to move the x-ray source and the detector relative to a patient as the gantry rotates around the patient so that the x-ray source and the detector follow a sinusoidal scan trajectory around the patient.

Further embodiments include an external beam radiation treatment system that includes a rotatable gantry having a head that emits a radiation treatment beam, the rotatable gantry configured to rotate around a patient to direct the radiation treatment beam to a target location within the patient from different angles, an x-ray imaging system mounted to the rotatable gantry, the x-ray imaging system including an x-ray source, a detector opposite the x-ray source, the detector including a two-dimensional array of pixels having a pixel spacing between adjacent pixels, and an apparatus for moving the detector array relative to the x-ray source by a distance that is less than the pixel spacing, and a processing device, coupled to the detector, that is configured with processor-executable instructions to perform operations including receiving, from the detector, a plurality of first x-ray images of the patient with a first spatial resolution that are obtained while the detector array is moved relative to the x-ray source by a distance that is less than the pixel spacing, and generating at least one second image of the object using the plurality of first x-ray images, wherein the at least one second image is a super resolution (SR) image that has an improved spatial resolution and/or signal-to-noise (SNR) ratio compared to the first images.

Further embodiments include an external beam radiation treatment system that includes a rotatable gantry having a head that emits a radiation treatment beam, the rotatable gantry configured to rotate around a patient to direct the radiation treatment beam to a target location within the patient from different angles, an x-ray imaging system mounted to the rotatable gantry, the x-ray imaging system including an x-ray source, a detector opposite the x-ray source, a first actuator system for moving the x-ray source relative to the rotatable gantry, and a second actuator system for moving the detector relative to the rotatable gantry, and a control system, coupled to the gantry, the first actuator system and the second actuator system, and including a processor that is configured with processor-executable instructions to perform operations including controlling the gantry to rotate the x-ray source and the x-ray detector around the patient in a first rotational direction and then in a second rotational direction that is opposite the first direction, controlling the first actuator system and the second actuator system to translate the x-ray source and detector system in a first translation direction along the length of the patient as the x-ray source and detector system rotate around the patient in the first rotational direction and the second rotational direction so that the x-ray source and detector system follow a reverse helical scan trajectory around the patient.

BRIEF DESCRIPTION OF THE DRAWINGS

Other features and advantages of the present invention will be apparent from the following detailed description of the invention, taken in conjunction with the accompanying drawings of which:

FIG. **1** illustrates a mobile x-ray imaging system.

FIG. **2** illustrates a multi-axis x-ray imaging system having a cantilevered imaging gantry.

FIG. **3** illustrates components mounted to a rotor of an x-ray imaging system.

FIGS. **4**A-**4**G illustrate a detector system for an x-ray imaging system according to various embodiments.

FIGS. **5**A-**5**C illustrate a hybrid x-ray imaging system for performing fan- and cone-beam CT imaging and 2D fluoroscopy.

FIG. **6** is a process flow diagram illustrating a method for obtaining x-ray images with improved spatial resolution using super resolution.

FIGS. **7**A-**7**G schematically illustrate a method of performing cone beam CT imaging using a sinusoidal circular trajectory.

FIGS. **8**A-**8**B schematically illustrate an x-ray imaging gantry rotated about a vertical axis for performing cone beam CT imaging using a sinusoidal spherical trajectory.

FIG. **9** illustrates an external beam radiation treatment system that includes an integrated x-ray imaging system.

FIG. **10** schematically illustrates an external beam radiation treatment system with an integrated x-ray imaging system having a detector that is moveable with respect to the focal spot of the x-ray source to increase the field-of-view of the imaging system.

FIG. **11** schematically illustrates an external beam radiation treatment system with an integrated x-ray imaging system that performs cone beam CT imaging using a sinusoidal scan trajectory.

FIG. **12** schematically illustrates an external beam radiation treatment system with an integrated x-ray imaging system that performs cone beam CT imaging using a reverse helical trajectory.

FIG. **13** illustrates a mobile x-ray imaging device that may be used to perform various embodiment methods.

FIG. **14** schematically illustrates a computing device which may be used for performing various embodiments.

DETAILED DESCRIPTION

The various embodiments will be described in detail with reference to the accompanying drawings. Wherever possible, the same reference numbers will be used throughout the drawings to refer to the same or like parts. References made to particular examples and implementations are for illustrative purposes, and are not intended to limit the scope of the invention or the claims.

An imaging system **100** according to one embodiment is shown in FIG. **1**. The imaging system **100** may be a mobile imaging system **100**, and may include a base **20**, a gimbal **30** and a gantry **40**. The system **100** includes image collection and generation components, such as an x-ray source and an x-ray detector system that are housed within the gantry **40**. The gantry **40** may be a generally O-shaped structure having a central imaging bore **116** and defining an imaging axis **114** extending through the bore. The system **100** is configured to collect imaging data, such as 3D x-ray tomographic image data and/or 2D x-ray fluoroscopic images, for example, from an object (e.g., human or animal patient) located within the bore **116** of the gantry **40**. In the embodiment of FIG. **1**, a column **50** is attached to the base **20** and supports a table **60** (e.g., a patient table) in a cantilevered manner such that the table **60** extends over the base **20** and at least partially into the bore **116** of the gantry **40**. It will be understood that in other embodiments, the column **50** and table **60** may not be attached to the base **20**.

The gimbal **30** may be a generally U-shaped support that is mounted to the top surface of base **20** and includes a pair of arms **31**, **33** extending up from the base. The arms **31**, **33** may be connected to opposite sides of gantry **40** so that the gantry is suspended above the base **20** and gimbal **30**. A drive system **70** may drive the translation (z-axis translation) of the gimbal **30** and the gantry **40** relative to the base **20** in the direction of arrow **104** shown in FIG. **1**. The drive system **70** may be mounted beneath the gimbal **30** within an open region of the base **20**, as shown in FIG. **1**. A control system **113** (e.g., a processor and memory) may be operatively coupled to the drive system **70** and may control the translation of the gimbal **30** and gantry **40** along the base **20** when the imaging system **100** is in a scan position (i.e., lowered to the floor). The control system **113** may be located on the imaging system **100**, such as on the gimbal **30**. A pair of rails **23** located on the top surface of the base **20** may engage with bearing elements on the bottom of the gimbal **30** to guide the translation of the gimbal **30** and gantry **40** along the base **20**.

The system **100** of FIG. **1** is a mobile system in which the entire system (including the base **20**, gimbal **30** and gantry **40**) may be moved across the floor (e.g., via one or more wheels or casters) for transport and/or repositioning of the system. For example, the base **20** may be raised from the floor while a plurality of casters are extended from the bottom of the base **20** and the entire system **100** may be driven in a transport mode. To perform an imaging procedure, the base **20** may be lowered to the ground and the gimbal **30** and gantry **40** may be translated along the length of the base **20** (e.g., to perform a CT scan). In the embodiment of FIG. **1**, the drive mechanism for transporting the entire system and the drive mechanism for translating the gimbal **30** and gantry **40** may be separate drive systems mounted beneath the gimbal **30**, which are collectively illustrated as drive system **70** in FIG. **1**. Various embodiments of a drive system **70** for a mobile imaging system as shown in FIG. 1 are described in U.S. Pat. No. 8,753,009 and U.S. Published Patent Application No. 2014/0265182, which are incorporated by reference herein.

The gantry **40** and at least an upper portion of the gimbal **30** may rotate on a rotary bearing **103** with respect to the base **20**. The rotation of the gantry **40** and gimbal **30** may be about a vertically-extending axis **102** in the direction of arrow **105** in FIG. **1**. The axis **102** may extend through the isocenter of the imaging gantry **40**. In some embodiments, a motorized system **112** as shown schematically in FIG. **1** may drive the rotation of the gantry **40** and gimbal **30** about axis **102**. The motorized system **112** may be operatively coupled to and controlled by a system controller **113**.

The gantry **40** may rotate (i.e., tilt) with respect to the gimbal **30** on a pair of rotary bearings **106** that attach the gantry **40** to the arms **31**, **33** of the gimbal **30**. The tilting motion of the gantry **40** may be about a horizontally-extending axis **107** in the direction of arrow **108** in FIG. **1**. The axis **107** may extend through the isocenter of the imaging gantry **40**. In some embodiments, a motorized system **115** shown schematically in FIG. **1** may drive the tilt motion of the gantry **40** relative to the gimbal **30**. The motorized system **115** may be operatively coupled to and controlled by a system controller **113**.

FIG. **2** illustrates another imaging system **200** according to an embodiment. The system **200** includes a gantry **40** that houses image collection components, such as a rotating x-ray source and detector array, that may be similar to the gantry **40** in the system **100** of FIG. **1**. The gantry **40** in system **200** may be mounted to a support column **201**. The support column **201** may be attached to the gantry **40** on a first side of the gantry **40** and may support the gantry **40** in a cantilevered manner. The gantry **40** may be a generally O-shaped structure having a central imaging bore **116** and defining an imaging axis **114** extending through the bore. The system **200** may also include a base **202** that may be located on a weight-bearing surface, such as a floor **219** of a building. In the illustrated embodiment, the base **202** comprises a generally rectilinear support structure that may be mounted (e.g., bolted) to the floor **219**. The support column **201** may be located on and supported by the base **202** and may extend upwards from the top surface of the base **202** in a generally vertical direction. The support column **201** may have a length dimension that extends vertically at least about 2 meters, such as 2-5 meters (e.g., about 3 meters).

In various embodiments, the system **200** may enable imaging (e.g., CT scanning, x-ray fluoroscopy) in multiple orientations and along multiple directions. In embodiments, the system **200** may include a first drive mechanism **203** for translating the gantry **40** relative to the support column **201** in a first direction along the direction of arrow **204** in FIG. **2**. The first direction **204** may be a generally vertical direction (i.e., perpendicular to the floor **219**), which for the purposes of this disclosure may be defined as ±15° from true vertical. The system **200** may also include a second drive mechanism **205** for rotating the gantry **40** relative to the support column **201** in the direction indicated by arrow **206**. The rotation of the gantry **40** may be with respect to an axis **207** that extends orthogonal to the first direction **204** and may be generally parallel to the floor **219**. The axis **207** may extend through the isocenter of the bore **116** of the gantry **40**. The system may also include a third drive mechanism **208** for translating the gantry **40** and support column **201** with respect to the base **202** in a second direction indicated by arrow **209** in FIG. **2**. The second direction **209** may be a generally horizontal direction (i.e., parallel to the floor **219**), which for the purposes of this disclosure may be defined as ±15° from true horizontal. The second direction **209** may be orthogonal to both the first direction **204** and to the rotation axis **207**.

In some embodiments, the gantry **40** may also pivot relative to the base **202**. For example, at least a portion of the support column **201** to which the gantry **40** is attached may rotate with respect to the base **202** in the direction of arrow **210** in FIG. **2**. A fourth drive mechanism **212** may drive the rotation of the support column **201**. The rotation of the support column **201** may cause the gantry **40** to pivot on the base **202**. The pivoting of the gantry **40** may be about a pivot axis **211** as shown in FIG. **2**.

The first drive mechanism **203**, the second drive mechanism **205**, the third drive mechanism **208** and/or the fourth drive mechanism **212** may be operatively coupled to and controlled by a system controller **213** that may be similar to the system controller **113** of FIG. **1**.

The imaging system **200** of FIG. **2** may enable imaging, including CT scanning and x-ray fluoroscopic imaging, of a human or animal patient in a horizontal orientation (e.g., lying on a patient table **60** as shown in FIG. **1**), in a vertical orientation (i.e., in a standing, weight-bearing position), or in a tilted orientation between a horizontal and vertical orientation. For example, the system **200** may perform a horizontal imaging scan (e.g., a helical CT scan) by translating the gantry **40** and support column **201** along the length of the base **202** with the imaging axis **114** extending horizontally along the length of a patient lying on a table (not shown for clarity). The system **200** may perform a vertical imaging scan by translating the gantry **40** along the length of the support column **201** with the imaging axis **114** extending vertically along the length of a patient supported in a weight-bearing position on a support structure (not shown) or standing directly on the floor **219**. The system **200** may also perform an imaging scan along an angled or oblique (i.e., neither vertical or horizontal) direction. Such a scan may be performed by rotating the gantry **40** on the support column **201** so that the imaging axis extends along the desired angle while performing coordinated translations of the gantry **40** along the length of the support column **201** and the support column **201** and gantry **40** along the length of the base **202**. Embodiments of a cantilevered multi-axis imaging system **200** such as shown in FIG. 2 are described in U.S. patent application Ser. No. 15/817,672, filed on Nov. 20, 2017, the contents of which are incorporated by reference herein. Embodiments of a patient table that may be used with a cantilevered multi-axis imaging system **200** for performing vertical, horizontal and/or angled imaging scans are described in U.S. Patent Application Publication No. 2018/0055707, the contents of which are incorporated by reference herein.

The gantry **40** in systems **100** and **200** may include an outer shell comprised of a high-strength structural material, such as aluminum. A rotor may be attached to the gantry shell via a bearing assembly that enables the rotor to rotate within the gantry **40**. FIG. **3** illustrates a rotor **41** for an x-ray CT imaging system having a plurality of components mounted thereto. The rotor **41** shown in FIG. **3** includes an x-ray source **43**, a high-voltage generator **44**, a heat exchanger **330**, an x-ray detector **45**, a power supply **63** (e.g., battery system), a computer **46**, a rotor drive mechanism **47**, and a docking system **35** (e.g., for providing intermittent power/data connection between components mounted to the rotor **41** and non-rotating portions of the system). It will be understood that the components described and illustrated are merely exemplary, and other embodiments may omit one or more of these components and may utilize other additional components. For example, in embodiments, power for the components on the rotor **41** may be provided by a slip ring or cable system, so that a power supply **63** on the rotor may not be needed. In some embodiments, power and/or data may be continuously transferred between rotating and non-rotating portions of the system via a cable, slip ring and/or wirelessly, in which case the power supply **63**, computer **46** and/or docking system **35** may not be included. Further, the rotation of the rotor **41** may be provided by a drive system on the non-rotating portion, in which case the rotor drive mechanism **47** on the rotor **41** may not be included.

In embodiments, the x-ray source **43** and detector **45** may be configured to perform a helical x-ray CT scan. The detector **45** may comprise a plurality of x-ray sensitive detector elements arranged in a generally semicircular arc, with the arc center coinciding with the focal spot of the x-ray source. Alternately, the x-ray detector may be a flat panel detector, and the system may be configured to perform real time x-ray fluoroscopic and/or cone beam imaging of an object within the bore of the gantry.

In the embodiment of FIG. **3**, during an imaging scan, the rotor **41** rotates within the interior of the gantry **40**, while the imaging components such as the x-ray source **43** and x-ray detector **45** obtain imaging data for an object positioned within the bore **116** of the gantry. The rotor drive mechanism **47** may drive the rotation of the rotor **41** around the interior of the gantry **40**. In embodiments, the rotor drive mechanism **47** may include a drive wheel **301** that engages with a belt **303**. The belt **303** may extend around the gantry **40** on a circular rail **305** that may be fixed to an interior wall of the gantry shell. The rotor drive mechanism **47** may be controlled by a system controller that controls the rotation and precise angular position of the rotor **41** with respect to the gantry **40**, preferably using position feedback data, such as from an encoder device. The rotation of the rotor **41** around the imaging bore **116** may be coordinated with a motion of the gantry **40** relative to the patient (e.g., a translation of the gantry **40** along the length of the patient using a system **100**, **200** as described above) to perform an imaging scan of a region of interest of the patient. Image data from the detector system **45** may be provided to a processing system (e.g., computer **46**) for generating a tomographic reconstruction of the region of interest.

FIGS. **4A-4G** illustrate an embodiment of a detector system **45** that may be utilized in an x-ray imaging system, such as the systems **100**, **200** described above in connection with FIGS. **1-3**. The detector system **45** may include a chassis **401** that may be mounted within the gantry **40** of the imaging system, and in particular, may be mounted to the rotor **41** opposite an x-ray source **43**, as shown in FIG. **3**. The chassis **401** may comprise a rigid frame comprised of a suitable structural material (e.g., aluminum), and may include a pair of parallel side walls **402**, **403** and a pair of end walls **404**, **405** defining an interior housing **406** (see FIG. **4B**) of the chassis **401**. The chassis **401** may have a curved or angled shape along its length dimension, as shown in FIGS. **4A-4B**. In embodiments, the chassis **401** may have a length of up to about 1 meter or more (e.g., 1.0-1.5 meters, such as 1.1 meter), a width of at least about 15 cm (e.g., 15-40 cm, such as ~20 cm), and a depth between a front side **407** and a rear side **409** of the chassis **401** that may be between about 10-15 cm.

A plurality of x-ray sensitive detector elements may be located within the interior housing **406** of the chassis **401**. In various embodiments, the individual detector elements may be located on a plurality of detector modules **410** (see FIGS. **4C-4F**). Each individual detector element **411**, which may be for example, a scintillator material coupled to a photodiode, represents a pixel on a multi-element detector module **410**. The modules **410** may be 2D element array, with for example 640 pixels per module (e.g., 32×20 pixels), although a greater or lesser number of pixels may be located on a module **410**. The pixel pitch of a module **410** may be less than 2 mm, such as about 1 mm (e.g., 0.7-1.3 mm). The detector elements **411** may be arranged in a planar array on the module **410**. In other embodiments, the detector elements **411** may be arranged in a non-planar (e.g., curved) array on the module. Each module **410** in a detector system **45** may have a uniform size and shape or the modules **410** may have varying sizes and/or shapes.

FIGS. **4E** and **4F** illustrate a pair of detector modules **410**. Each module **410** includes a layer of scintillator material **412**, such as gadolinium oxysulfide (GOS), a photodiode array **413** optically coupled to the scintillator material **412**, and a front-end electronics assembly **414** coupled directly behind photodiode array. The electronics assembly **414** may include analog-to-digital (A/D) converter circuitry for converting the output signals from the photodiode array to digital signals. The signals may comprise a digital representation of an intensity of x-ray photons incident on each individual detector element **411** (i.e., pixel) of the array. The electronics assembly **414** may be closely coupled to the photodiode array in order to minimize signal path length and signal path length differences among the pixels of the detector module **410**. This may help minimize noise and increase SNR of the detector module **410**. The electronics assembly **414** may be enclosed within a protective housing **415** as shown in FIGS. **4E** and **4F**.

A plurality of modules **410** may be arranged in an abutting fashion to provide a detector area with a desired size and shape. For example, a plurality of modules **410** may be abutted along the width dimension of the detector system **45** (i.e., parallel to the imaging axis **114** of the gantry **40**) to provide an arbitrary number of "slices" that may be imaged simultaneously during a rotation of the rotor **41**. The number of slices of the detector system **45** may be, for example, 32 slices, 40 slices, 64 slices, 128-320 slices, etc. In the embodiment shown in FIGS. **4**E and **4**F, two detector modules **410**, each 32 pixels in width, are abutted side-by-side within the detector chassis **401** to provide a 64 slice CT detector system **45**. The gap between adjacent modules **410** in the width direction may be less than 0.5 mm, such as less than 0.05 mm (e.g., 0.001-0.05 mm), and may be, for example, between 0.01 and 0.02 mm.

The modules **410** may also be abutted in the transverse direction (i.e., along the length of the chassis **401**) to provide a detector area having an arbitrary length. The gap between adjacent modules in the length direction may be less than 1 mm, such as less than 0.5 mm (e.g., 0.01-0.5 mm), and may be, for example, between 0.4 and 0.5 mm. In the detector system **45** of FIGS. **4**A-**4**G, the detector system **45** may include two adjacent rows of detector modules **410**, each row being 55 modules in length, for a total of 110 modules. The modules **410** may be curved or angled along the length of the detector chassis **401** to form or approximate a semi-circular arc, with the arc center coinciding with the focal spot **307** of the x-ray source **43** (see FIG. **3**). For example, each of the modules **410** may comprise a planar, 2D pixel array that is mounted within the housing **406** of the chassis **401** such that a ray extending from the focal spot **307** of the x-ray source **43** is perpendicular to the surface of the module **410** at the center pixel of the module **410**.

The detector system **45** may include a plurality of sub-assemblies **416**, each including one or more detector modules **410**, where the sub-assemblies **416** may be individually mounted within and removed from the detector chassis **401**. FIG. **4**B is a rear view of a detector system **45** showing a plurality of sub-assemblies **416** mounted to the detector chassis **401**. FIG. **4**C is a cross-section view of the detector system **45** (viewed along the width dimension of the detector system **45**) showing a sub-assembly **416** mounted to the detector chassis **401**. FIG. **4**D is another partial cross-section view of the detector system **45** (viewed along the length dimension of the detector system **45**) showing a pair of adjacent sub-assemblies **416** mounted to the detector chassis **401**. FIGS. **4**E and **4**F are perspective views of a sub-assembly **416**. The sub-assembly **416** in this embodiment includes a support **417** that is configured to receive a pair of adjacent detector modules **410** on a first side of the support **417**. The support **417** may comprise a structural material (e.g., aluminum) that preferably has a relatively high thermal conductivity (e.g., ≥200 W/(m·K)). The support **417** may function as a heat sink to facilitate removal of heat from the modules **410**, and in particular from the electronics assembly **414** of the modules **410**. In embodiments, a thermal paste may be provided at the interface between the detector modules **410** and the support **417**. The detector modules **410** and the support **417** may have alignment features that enable the precise positioning of the modules **410** on the support. For example, the modules **410** may include one or more pins that may be received in precision-formed hole(s) in the mating surface of the support **417**. Although the sub-assembly **416** shown in FIGS. **4**E and **4**F supports two detector modules **410**, it will be understood that a sub-assembly **416** may support a single detector module **410** or more than two detector modules **410**. The sub-assembly **416** may support a plurality of modules **410** such that they form a single planar surface, such as shown in FIGS. **4**E and **4**F, or may support the modules **410** such that they are angled relative to one another (e.g., such as to approximate an arc centered on the focal spot **307** of the x-ray source **43**).

The sub-assembly **416** may also include an anti-scatter assembly **418** attached to the support **417** over the detector modules **410**. The anti-scatter assembly **418** may comprise a grid **419** made from an x-ray absorbent material, such as tungsten or lead, having openings aligned over the pixels of the detector modules **410**. The x-ray absorbent material of the grid **419** may be located between adjacent columns and rows of detector elements (pixels) of the module **410** so as to inhibit scattered x-rays, traveling at an oblique angle, from impinging on the detector elements. The anti-scatter assembly **418** may be provided so that the detector elements primarily measure the intensity of x-rays that travel along a straight-line path from the x-ray source **43** to the detector **45**. The anti-scatter assembly **418** may comprise a focused grid having apertures that are angled toward the focal spot of the x-ray source. Although the anti-scatter assembly **418** in this embodiment is a two-dimensional grid **419** located between each column and row of detector elements, it will be understood that other configurations can be utilized, including an array of x-ray absorbent plates located between adjacent columns or rows of detector elements to absorb scattered x-rays along one direction.

The sub-assembly **416** may also include at least one electronics board **420** attached to a second side of the support **417** that is opposite the first side to which the module(s) **410** are attached. One or more electronic cables **421** may extend between each of the modules **410** and the electronics board **420**. As shown in FIGS. **4**E and **4**F, a pair of ribbon cables **421** extends between the electronics assemblies **414** of the modules **410** and the electronics board **420**. The cables **421** may extend in recessed portions **443** (e.g., grooves) extending along the side of the support **417** that are sized and shaped to accommodate the cables **421**, as shown in FIGS. **4**E and **4**F. The cables **421** may provide power from the electronics board **420** to each of the modules **410**. The cables **421** may also carry image data (e.g., digital x-ray attenuation data) from the modules **410** to the electronics board **420**. The electronics board **420** may include memory and at least one processor (e.g., an FPGA) programmed to receive digital image data from the electronics assemblies **414** of the detector modules **410**, and assemble the data into a single image. The detector system **45** may include a "double buffering" configuration, such that while a first plurality (e.g., frame) of image data accumulates in a first buffer, a second plurality (e.g., frame) of digital image data may be read out from a second buffer. The data from the first buffer may then be read out while new data accumulates in the second buffer, and so on. The detector modules **410** may continually collect digital image data while the detector **45** is continuously exposed by the x-ray source **43**. In some embodiments, the electronics assemblies **414** of the detector modules **410** may transmit the image data in a digital video format, such as LVDS. The at least one processor on the electronics board **420** may be configured to convert the image data to a different digital video format, such as Camera Link or gigabit Ethernet.

The electronics board **420** may be secured to the support **417** of the sub-assembly **416** using an adhesive and/or suitable fastener(s). A handle **422** or other element to enable the sub-assembly **416** to be easily gripped and manipulated may be located over the electronics board **420** opposite the support **417**. One or more fasteners (e.g., screws) may extend through the handle **422** and the electronics board **420** and into the support **417** to clamp the electronics board **420** to the support **417**.

As shown in FIGS. **4**A and **4**C, the front side **407** of the detector chassis **401** may include an open window region **424** over the detector modules **410**. A cover **425** made of a material that is x-ray translucent but opaque to visible radiation (e.g., a thermoplastic material, carbon fiber, etc.) may be mounted to the front side **407** of the detector chassis **401** over the plurality of detector modules **410**. The rear side **409** of the detector chassis **401** may include a plurality of access openings **426**, as shown in FIG. **4**B. A plurality of access panels may be secured over the openings **426**. FIG. **4**C illustrates an access panel **429** in phantom. The access panels **429** may include interlocking features to prevent ambient light from entering the detector system **45**.

As shown in FIGS. **4**B, **4**C and **4**D, a pair of lip portions **430** may extend along the interior sidewalls of the housing **406** of the detector chassis **401**. The lip portions **430** may have a curved or faceted profile over the length of the chassis **401**, as shown in FIG. **4**B, so that when the detector sub-assemblies **416** are mounted to the chassis **401** the detector modules **410** are arranged in a generally semicircular arc configuration along the length of the chassis **401**. Each of the detector sub-assemblies **416** may include one or more alignment pins **431** (see FIGS. **4**E and **4**F) for aligning the sub-assembly **416** within the detector chassis **401**. The alignment pins **431** may be received in openings **432** on the lip portion(s) **430** of the chassis **401** (see FIG. **4**B). A set of fasteners **433** (e.g., screws) may be inserted through the support **417** and into the respective lip portions **430** to secure the sub-assembly **416** to the chassis **401**. A plurality of sub-assemblies **416** may be mounted to the lip portions **430** of the detector chassis **401**. FIG. **4**B illustrates six sub-assemblies **416** mounted to the chassis **401**. It will be understood that a detector system **45** may include a series of sub-assemblies **416** (e.g., 55 sub-assemblies) mounted adjacent to one another along the length of the chassis **401**.

The support **417** of the sub-assembly **416** may also include a set of alignment pins **434** to facilitate alignment of the anti-scatter assembly **418** over the detector modules **410**. The alignment pins **434** may be received in a recessed portion **435** of the anti-scatter assembly **418**, as shown in FIGS. **4**E and **4**F. A set of fasteners **436** (e.g., screws) may be inserted through the anti-scatter assembly **418** and into the support **417**.

In embodiments, to access the detector modules **410** and/or electronics boards **420** for service and/or replacement, one or more access panels **429** may be removed from the rear side **409** of the detector chassis **401**, and an individual sub-assembly **416** may be disconnected and removed from the detector chassis **401** without disturbing other sub-assemblies **416** of the detector system **45**.

In the embodiment shown in FIGS. **4**B-**4**F, the electronics board **420** of each sub-assembly **416** may include a cantilevered end **437** that projects out from the side of the sub-assembly **416**. The cantilevered end **437** may include a connector **438** (e.g., plug) that connects to an additional circuit board **439** (e.g., a motherboard) extending on the periphery of the detector system **45** (e.g., between the lip portion **430** and the side wall **403** of the detector chassis **401**). Alternately, the electronics board **420** of the sub-assembly **416** may connect to the additional circuit board **439** via a cable connection (e.g., a ribbon cable). The additional circuit board **439** may extend perpendicular to the electronics board **420** and parallel to side wall **403** of the detector chassis **401**. The additional circuit board **439** may be connected to a plurality of sub-assemblies **416** (e.g., at least two sub-assemblies **416**, and in some embodiments, all of the sub-assemblies **416** of the detector **45**). Connectors **438** between each of the electronics boards **420** and the additional circuit board **439** may be utilized to provide power to the electronics boards **420** and to receive data (e.g., digital x-ray data) from each of the electronics boards **420**. The additional circuit board **439** may include memory and at least one processor (e.g., an FPGA) programmed to receive digital image data from the sub-assemblies **416** and to combine and assemble the data into a single image. In embodiments, electronics boards **420** may be data acquisition system (DAS) boards that may provide circuit integration and the transfer of data from the electronics modules **410** to a motherboard (e.g., board **439**).

The detector system may include a plurality of additional circuit boards **439** that may extend along the length of the detector chassis **401**. FIG. **4**B illustrates four circuit boards **439** extending along the length of the chassis **401**, although it will be understood that a greater or lesser number of boards may be utilized. Each of the additional circuit boards **439** may receive and combine digital image data from a group of sub-assemblies **416** to generate a combined dataset. The additional circuit boards **439** may be connected to one another in a daisy-chain fashion such that each board **439** may combine its own image dataset with an image dataset received from a preceding board **439** in the chain. The resulting combined dataset may then be transmitted to a subsequent board **439** which may repeat this process so that the image data is propagated down the line to a final board of the chain (e.g., board **439**A in FIG. **4**B). The final board **439**A may include one or more external connectors or ports **440** (see FIG. **4**A) for receiving power from a suitable power supply (e.g., a battery system **63** as shown in FIG. **3**) and for transmitting the combined image dataset to an external device (e.g., computer **46** as shown in FIG. **3**) for performing processing operations, including tomographic reconstruction. FIG. **4**G is an enlarged view of an interface between the detector system **45** and a plurality of external connectors **440**. As shown in the enlarged view of FIG. **4**G, the external connectors **440** may plug directly into a circuit board **439**A of the detector system **45**. The side wall **403** of the detector chassis **401** may include an opening **441** to enable the connector(s) **440** to connect to the circuit board **439**A. A compliant member **442** (e.g., gasket) may surround the opening **441** between the chassis side wall **403** and the circuit board **439**A.

In embodiments, the detector chassis **401**, including at least the side walls **402**, **403**, end walls **404**, **405** and lip portions **430** to which the detector sub-assemblies **416** are mounted, may be formed (e.g., precision machined) from a single workpiece (e.g., an aluminum block). This may improve the manufacturability of the detector system while ensuring that required tolerances are met.

The detector chassis **401** may be made from a material having a relatively high thermal conductivity (e.g., ≥200 W/(m·K)), such as aluminum. Cooling of the interior housing **406** of the detector chassis **401** may be primarily through conduction. As discussed above, each of the sub-assemblies **416** of the detector may include a support **417** that functions as a heat sink to conduct heat away from the detector modules **410** and other electrical components. The sub-assemblies **416** are mechanically and thermally coupled to the detector chassis **401** (i.e., at the interface between the support **417** and the lip portions **430** of the chassis **401**) so that heat is conducted from the sub-assemblies **416** into the chassis **401**. The supports of the sub-assemblies and the chassis walls may provide a thermally-conductive path from the detector modules **410** and electronics boards **420** to the exterior surfaces of the chassis **401**. In embodiments, a thermal paste may be provided at the interface between the sub-assemblies **416** and the chassis **401**. Other heat generating components of the detector system **45**, such as a power supply and the additional circuit board(s) **439**, may be directly mechanically and thermally coupled to the chassis **401**, such as fastened to the interior side wall **403** of the chassis **401**. A cooling fluid (e.g., air) may be directed through the interior of the gantry **40** and over the exterior surface of the detector chassis **401** to cool the detector system **45**.

Alternately or in addition, the detector system **45** may include a cooling system for directing a cooling fluid (e.g., air) through the interior housing **406** of the detector chassis **401** to remove heat, such as described in U.S. Pat. No. 9,125,613, the entire contents of which are incorporated by reference herein.

The heat load of the detector system **45** may also be managed by controlling the power provided to the detector system **45** (e.g., from batteries **63**) so that all or a portion of the electrical components of the detector system **45** may remain unpowered or minimally powered when not in use. A control system (e.g., controller **113**) may maintain the detector system **45** in an unpowered or low-power "standby" mode until the imaging system is ready to perform an imaging scan. Upon initiation of a scan, the control system may transmit a control signal to cause the detector system **45** to be powered up to an operational mode in which the detector system **45** is ready to obtain x-ray image data. After the scan is complete, the detector system **45** may be returned to an unpowered or low-power standby mode until the next scan is initiated.

In an alternative configuration of the detector system **45**, the electronics boards **420** of the sub-assemblies **416** may be directly connected to one another, such as via a plurality of ribbon cables. Additional circuit board(s) **439** extending along the length of the chassis **401** may optionally be omitted. The electronics boards **420** of the plurality of sub-assemblies **416** may transmit the collected image data along a series of adjacent sub-assemblies **416** in a daisy chain configuration, such as described in U.S. Pat. No. 9,111,379, the entire contents of which are incorporated by reference herein. Each sub-assembly **416** may combine its own image data with the image data received from a preceding sub-assembly **416**, and transmit the combined image data to the next sub-assembly **416** in the series. The combined image dataset from all of the sub-assemblies **416** may be streamed into a frame grabber (e.g., where the data may be converted to gigabit Ethernet format) and the frames of data may be streamed into an external processing device (e.g., computer **46**) for performing tomographic reconstruction and/or other processing operations.

In some embodiments, at least a portion of the image data processing operations, including tomographic reconstruction, may be performed within the detector system **45** itself. In embodiments, at least one image processing/reconstructor module may be located within the detector system **45**. The image processing/reconstructor module may be implemented on a separate computer located on or within the detector chassis **401** and/or on one or more circuit boards **439** within the detector chassis **401**. The image processing/reconstructor unit may comprise a parallel processor having a plurality of processing cores for performing the tomographic reconstruction process in parallel. For example, the parallel processor may be a graphics processing unit (GPU), and may be located on a graphics card. The GPU may include a large internal memory (e.g., up to 8 gigabytes or more, such as 2-4 gigabytes) and a plurality of processing cores (e.g., up to 4096 cores or more, such as 2048 cores) for performing parallel processing of the image data. It will be understood that the image processing/reconstructor module may be implemented using any suitable processing device, such as one or more of a GPU, a CPU, an FPGA, ASIC, etc. The image processing/reconstructor module may receive inputs of encoder position(s) (e.g., indicating the rotation position of the rotor **41** and/or translation/rotation of the gantry **40**), source-to-detector distance, x-ray photon flux and/or x-ray source temperature (e.g., from a reference detector as described in U.S. Pat. No. 9,111,379). The image processing/reconstructor module may be configured to perform various image correction techniques on the image data, such as offset correction, gain correction and/or pixel correction. In some embodiments, the image processing/reconstructor module may also perform other real-time processing operations (e.g., for 2D fluoroscopy), including edge enhancement, recursive noise reduction and super resolution techniques as described below.

FIGS. **5A-5C** illustrate another embodiment imaging system **500** that may be used for both 3D CT imaging and 2D fluoroscopic imaging. FIG. **5A** is a cross-section view of a gantry **40** of an imaging system **500**, which may be similar to systems **100**, **200** shown in FIGS. **1** and **2**. The gantry **40** includes a rotor **41** that is mounted to and rotates within an outer shell **42** of the gantry **40**. A plurality of components, including an x-ray source **43**, a high-voltage generator **44**, a heat exchanger **330**, an x-ray detector system **501**, a power supply **63** (e.g., battery system), a computer **46**, a rotor drive mechanism **47**, and a docking system **35**, may be mounted to the rotor **41** as described above.

The detector system **501** in this embodiment is a hybrid detector system having a contiguous detector area that includes an elongated first portion **502** for performing fan-beam CT imaging (e.g., axial and/or helical scans) and a panel-shaped second portion **503** for performing 2D fluoroscopic imaging and/or 3D cone beam CT imaging. The first portion **502** and the second portion **503** may be overlapping, such that a portion of the detector area is shared by both the first portion **502** and the second portion **503**. The first portion **502** may have a length dimension (L**1**) that is greater than the length dimension (L**2**) of the second portion **503**. For example, the first portion **502** may have a length that is greater than 0.5 meter, such as 1 meter or more, and the second portion **503** may have a length that is less than 0.5 meter, such as between about 0.3 and 0.4 meters. The second portion **503** may have a width dimension (W**2**) that is greater than the width dimension (W**1**) of the first portion **502**. For example, the first portion **502** may have a width that is less than 0.3 meters (e.g., 0.15-0.25 meters) and the second portion **503** may have a width that is greater than 0.3 meters (e.g., 0.3-0.4 meters or more).

The detector area may be produced by arranging an array of detector modules **504** in a desired geometric shape or pattern. The modules **504** may be similar or identical to the modules **410** described above with reference to FIGS. **4A-4G**. For example, each module **504** may include an array of individual detector elements (pixels), each including a scintillator (e.g., GOS) coupled to a photodiode, and including an electronics assembly for outputting digital image data. The modules **504** may be abutted along any of their edges to form a detector area having any arbitrary size and shape. In the embodiment of FIGS. **5A-5C**, the first portion **502** of the detector area may be formed by abutting a group of modules **504** along the length dimension, L**1**, and the width dimension, W**1**. For example, the first portion **502** may include two adjacent rows of detector modules **504**, each row being 55 modules in length, for a total of 110 modules, similar to the detector system **45** shown in FIGS. **4**A-**4**G. The first portion **502** may be a large field-of-view (e.g., providing ~50 cm diameter or greater reconstruction volume), multi-slice (e.g., 64 slice) true CT detector.

The second portion **503** of the detector area may be formed by abutting additional row(s) of modules **504** in the width direction along a section of the modules **504** forming the first portion **502**. In the embodiment of FIGS. **5**A-**5**C, the second portion **503** of the detector area may be formed by abutting three additional rows of modules **504** on either side of a central section of the two rows of modules **504** forming the first portion **502**. The length of the central section may define the length dimension, L**2**, of the second portion **503**. The distance between the edges of the outer two rows of modules **504** in the second portion **503** may define the width dimension, W**2**, of the second portion **503**. In the embodiment of FIGS. **5**A-**5**C, the second portion **503** of the detector area includes eight adjacent rows of detector modules **504**, each row being nineteen modules in length, for a total of 152 modules. The second portion **503** may be a rectangular panel detector that may be used for 2D x-ray fluoroscopy and/or cone-beam CT imaging.

The detector modules **504** in the detector system **501** may have a uniform size and shape or may have varying size(s) and/or shape(s). In one embodiment, the modules **504** may be a 2D element array, with for example 640 pixels per module (e.g., 32×20 pixels). The modules **504** may be mounted within a housing of a detector chassis **505**, which may be similar to the chassis **401** shown in FIGS. **4**A-**4**D. In FIGS. **5**A-**5**C, the chassis **505** is schematically illustrated supporting the array of detector modules **504**. In embodiments, the chassis **505** may enclose the modules **504** within an internal light-tight housing (not illustrated in FIGS. **5**A-**5**C for clarity). The chassis **505** may be sized and shaped to accommodate the first and second portions **502**, **503** of the detector area.

The modules **504** may be supported on the chassis **505** such that the modules **504** are curved or angled along the length of the chassis **505** to form or approximate a semicircular arc, with the arc center coinciding with the focal spot **307** of the x-ray source **43** (see FIG. **3**). In some embodiments, the modules **504** of the second portion **503** may additionally be curved or angled along the width of the chassis **505** to form or approximate a semicircular arc centered on the focal spot **307** of the x-ray source **43**. The modules **504** of the second portion **503** may thus approximate a portion of a spherical surface that is centered on the focal spot **307** of the x-ray source **43**.

Each detector module **504** may be electronically coupled to an electronics board (e.g., similar or identical to the electronics board **420** shown in FIGS. **4**B-**4**G) that provides power to the module **504** and receives digital image data from the module **504**. The detector system **501** may include a plurality of electronics boards that are arranged behind the detector modules **504** within the chassis **505**. Each electronics board may be electrically connected to more than one detector module **504**. The electronics boards may have a uniform size and shape and may be connected to the same number of detector modules **504**. Alternately, the electronics boards may have different sizes and shapes and may be connected to different numbers of detector modules **504**. For example, the electronics boards of the first portion **502** of the detector system **501** may each connect to a first number of detector modules **504** (e.g., 2 modules, such as in the embodiment of FIGS. **4**A-**4**G), and the electronics boards located behind the peripheral rows of the second portion **503** of the detector system **501** may each connect to a different number of detector modules **504** (e.g., greater than 2 modules, such as 3-10 modules). Each of the electronics boards may send the image data received from the detector modules **504** to a processing unit (e.g., an FPGA) that may be configured with software to unscramble the data and assemble a combined image dataset (e.g., image frame) from all of the detector modules **504** in a selected area that is exposed to x-ray radiation.

In embodiments, the detector system **501** may include a plurality of sub-assemblies that may be individually mounted within and removed from the detector chassis **505**. The sub-assemblies may be similar or identical to the sub-assemblies **416** shown in FIGS. **4**B-**4**F and may include a support that is mounted to the detector chassis **505**, where the support includes one or more detector modules **504** and one or more electronics boards attached thereto. In one non-limiting example, a first group of sub-assemblies forming the elongated, narrower "wing" sections of the first portion **502** (i.e., that do not overlap with the second portion **503**) may be identical to the sub-assemblies **416** described above in connection with FIGS. **4**B-**4**F. A second group of sub-assemblies may be mounted within the chassis **505** to form the second portion **503** of the detector system **501**. Each of the second group of sub-assemblies may include a greater number of detector modules **504** (e.g., eight modules) attached to a support that extends across the width of the detector system **501**. Alternately, the chassis **505** may include a plurality of support ribs extending within the second portion **503** of the second portion **503** of the detector to which the sub-assemblies may be attached.

The detector system **501** may also include an anti-scatter assembly located over the detector modules **504**. The anti-scatter assembly may include a two-dimensional grid comprised of x-ray absorbent material located between the columns and rows of detector elements (pixels) or may be an array of x-ray absorbent plates located between adjacent columns or rows of detector elements. The anti-scatter assembly may include grid or plate elements mounted above the detector modules in each sub-assembly of the detector system **501**, as shown in FIGS. **4**C-**4**F.

In embodiments, the x-ray source **43** of the imaging system **500** may include an adjustable collimator **506** that defines the shape of the x-ray beam **517** emitted by the source **43**. The collimator **506** may include motor-driven shutters or leaves comprised of an x-ray absorbent material (e.g., lead or tungsten) that may block a portion of the x-rays generated by the x-ray tube. In a first configuration shown in FIG. **5**B, the collimator **506** may collimate the beam **517** so that it covers the first portion **502** of the detector area. In a second configuration shown in FIG. **5**C, the collimator **506** may collimate the beam **517** so that it covers the second portion **503** of the detector area. The first configuration shown in FIG. **5**B may be utilized, for example, for performing large field-of-view fan-beam helical or axial CT scans. The second configuration shown in FIG. **5**C may be utilized for performing 2D fluoroscopic imaging and/or 3D cone-beam CT scans. In embodiments, the configuration of the collimator **506** may be adjusted by a system controller, which may be implemented on a computer (e.g., computer **46**). The system controller may also send a configuration signal to the detector system **501** to indicate the detector modules **504** from which to read out image data based on the shape of the x-ray beam **517**. The imaging system **500** as shown in FIGS. 5A-5C may be used to perform diagnostic-quality CT scans (e.g., multi-slice large field-of-view axial and/or helical scans), 2D fluoroscopic imaging and/or 3D cone beam CT imaging using a single x-ray source **43**, high-voltage generator **44** and detector system **501**.

A detector system **501** such as shown in FIGS. **5A-5C** may have advantages over conventional flat-panel detectors that are used for fluoroscopy and cone-beam CT imaging. For example, the detector system **501** may enable continuous exposure to x-ray radiation, which is in contrast with some flat-panel detectors that require pulsing or strobing of the x-ray source and may have a relatively low duty cycle. A detector system **501** according to various embodiments may utilize a double- or multi-buffering system as described above, such that the detector system **501** may continuously collect new x-ray image data while reading out previously-collected image data. The detector system **501** may be used to collect, and optionally display, x-ray image data at a relatively high frame rate (e.g., >30 Hz, such as 60 Hz or greater, and in some cases more than 1000 Hz). An embodiment detector system **501** may also increase the efficiency of the imaging system **500**. For example, conventional flat panel detector systems may require high power (e.g., ≥100 kW) generators to produce the required voltage and current within the tube. The reason for this is that the relatively low duty cycle of the detector limits the total dose of x-rays that can be used for each image frame per unit of time. For example, in a pulsed x-ray system with a 30 Hz frame rate but only 300 msec of x-ray exposure per second, in order to obtain 300 mA-seconds of data, one needs to expose at approximately 1000 mA per pulse. However, with a continuous exposure system according to the various embodiments, the tube current may be significantly lower to obtain the equivalent milliamp-seconds (e.g., 300 mA to get 300 mA-seconds). This may enable lower power (e.g., ≤40 kW) generators to be used and can reduce the size and power requirements of the system **500**.

A detector system **501** as described above may also provide improved image quality over an equivalent flat panel detector by including a curved profile (in one or two dimensions) in which the detector elements or modules **504** are arranged along an arc centered on the x-ray focal spot, and may also include a 1D or 2D anti-scatter assembly as described above. The detector system **501** may enable diagnostic-quality cone-beam CT images over at least the center slice of the reconstruction. Further, a true CT detector system **501** as described above may have better dynamic range than a conventional flat panel detector used for fluoroscopic imaging.

In embodiments, the rotor **41** containing the x-ray source **43** and the detector system **501** may rotate within the gantry **40** at least 90° per second, including 180°, 270° or 360° or more per second. The rotation of the rotor **41** may be continuous (i.e., the rotor **41** may continuously rotate through 360° in the same direction). The rotation of the rotor **41** may be unidirectional or bi-directional (i.e., in both clockwise and counterclockwise directions). The system controller (e.g., computer **46**) may precisely control the rotation and rotational position of the rotor **41** by sending control signals to the rotor drive system **47**. The rotor position may be controlled based on encoder feedback data. The rotor components may be completely housed within the gantry **40** during their rotation to avoid any possibility of collision with a person or other object.

In embodiments, the rotor **41** may be controlled to move to different rotational positions to obtain x-ray fluoroscopic images from different projection angles (e.g., anterior-posterior, lateral images). The rotor **41** may rotate between the different projection angles in a short period of time (e.g., 500 msec or less) to provide updated "real time" fluoroscopic images from multiple projection angles.

The imaging system **500** may also be used to obtain 2D and/or 3D CT images along oblique projection angles by rotating/tilting the gantry **40** with respect to the patient (i.e., such that the patient axis is not parallel to the gantry imaging axis **114**). For example, a system that is configured as shown in FIG. **1** may enable isocentric rotation of the gantry **40** in a "tilt" direction (i.e., about axis **107**) and/or in a "wag" direction (i.e., about axis **102**). A system such as shown in FIG. **2** may enable isocentric "tilt" rotation of the gantry **40** (i.e., about axis **207**). Isocentric "wag" rotation may be achieved by performing a coordinated pivot motion of the gantry **40** and support column **201** (i.e., about axis **211**) and a translation of the gantry **40** and support column **201** (i.e., in the direction of arrow **209**) to maintain the isocenter of the gantry in the same location. In embodiments, an imaging system **500** as shown in FIGS. **5A-5C** may perform real-time x-ray fluoroscopy in wide number of projection angles, including steep oblique angles. The gantry **40** may also be translated along the patient (by a distance of up to at least about 1 meter) for performing fan-beam or cone-beam CT scans.

Further embodiments include methods and systems for improving the spatial resolution and/or signal-to-noise ratio (SNR) of x-ray images using super resolution (SR). In an imaging system such as the system **500** shown in FIGS. **5A-5C**, the spacing of the pixels (i.e., the center-to-center spacing of adjacent pixels in the detector system **501**) may define the spatial resolution of the imaging system. For example, a pixel spacing of 1 mm provides a special resolution of 1×1 mm at the detector face. This results in a resolution of about 0.5×0.5 mm at the isocenter of the imaging bore **116** of the system. Thus, a detector system **501** with a pixel spacing of ~1 mm may be used for 2D fluoroscopic or 3D tomographic images having a spatial resolution of about 0.5×0.5 mm.

There is an inherent tradeoff between pixel size and signal-to-noise ratio (SNR) in that the smaller the pixel size, the less photons are incident on each pixel per unit of time and the lower the SNR of the system. Thus, improving the spatial resolution by decreasing the pixel size may lower the SNR of the system below generally acceptable levels. This issue may be partially compensated by, for example, increasing the number of x-ray photons emitted by the source **43** (e.g., by employing a large, high-power (e.g., 120-130 kW) high-voltage generator coupled to the x-ray tube) and decreasing the exposure time (i.e., to maintain approximately the same x-ray radiation dose). However, these approaches may not be feasible due to size and power constraints of the system. Further, there may be an upper limit in terms of the ability of the system's electronics components (e.g., A/D converters, frame grabbers, etc.) to collect large quantities of raw projection data in a relatively short time period. In addition, the power of the x-ray source **43** can only be increased so much until radiation dosing and exposure levels become a safety issue. Thus, for all of these reasons there is generally a practical lower limit on the pixel size of the detector and therefore the achievable spatial resolution in the image.

Various embodiments include methods and systems for improving the spatial resolution of an x-ray imaging system using super-resolution. Embodiments may be implemented in software, and may improve spatial resolution of the image without requiring smaller pixel sizes for the detector. In addition, embodiments may improve the spatial resolution without decreasing the signal-to-noise ratio (SNR) of the detector. Further embodiments may improve the spatial resolution while maintaining or even decreasing the x-ray radiation dose received by the patient.

Various embodiments may improve the spatial resolution of the x-ray image in accordance with a factor, F, relative to the best spatial resolution that would otherwise be achievable in the image based on the size and/or spacing of the detector elements (i.e., pixels). As used herein, the spatial resolution of the image refers to the size of the smallest discernible feature of the image. Thus, in this context, "improving" the spatial resolution means that the spatial resolution is made smaller (i.e., such that smaller features are discernible in the image). By way of example, a detector having a pixel spacing, Sp may inherently result in a minimum achievable spatial resolution, R1, in the image without the use of an embodiment method. The various embodiments may improve (i.e., decrease) the achievable spatial resolution for the same detector in accordance with a factor, F, by providing an image having a second spatial resolution, R2, where R2=R1(1/F). In various embodiments, F>1.0, and may be between about 1.1 and about 10 (e.g., 1.2-5.0, such as 1.5-4.0), including between 1.5-2.0, 2.0-3.0, 3.0-4.0 and/or 4.0-5.0.

In one embodiment, for a detector having a pixel spacing Sp of about 1.0 mm or more (e.g., 1.0-2.0 mm), which would normally provide a spatial resolution of about 0.5 mm or greater in the image, the spatial resolution of the image obtained using the present invention may be improved to less than 0.5 mm (e.g., 0.1 mm to 0.45 mm, such as 0.2-0.4 mm, including about 0.25 mm).

The various embodiments may improve the resolution of the image using super-resolution. Super-resolution refers to a class of processing techniques for improving the resolution of an imaging system. In general, super-resolution techniques use information from several different images/frames to create an upsized image having improved resolution and lower noise (i.e., higher SNR). These techniques were originally developed in photography and video editing to improve the resolution beyond what would otherwise be possible based on the pixel size of the camera taking the images. Super-resolution techniques typically require that multiple images or frames be taken of the scene or object of interest from slightly different perspectives (e.g., the camera has moved slightly with respect to the object between images/frames). The techniques will not work if either the camera does not move at all relative to the object or the camera moves too much relative to the object between successive images/frames. In a typical super-resolution technique, an image/frame is upsized (i.e., the number of pixels is increased by some factor) and an interpolation algorithm is applied to generate an interpolated upsized image. Then, one or more regions within neighboring images/frames may be compared to estimate how much objects within the region(s) have moved between frames. Information from neighboring images/frames may then be intelligently merged with the interpolated upsized image to produce a super-resolution (SR) image which may have more information than is contained in any one of the originally-obtained images. In particular, the super-resolution image may have improved spatial resolution and/or SNR compared to the original image(s). The process may be repeated iteratively to further improve the spatial resolution and/or SNR.

In general, super-resolution techniques may involve attempting to reproduce or model the process by which image quality is lost using the camera/detector that obtained the lower-quality images and then solving the inverse problem of finding (e.g., reconstructing) the higher-quality image (e.g., an upsized image with improved spatial resolution and higher SNR) which would produce the known lower-quality images by that process. Various techniques and algorithms for improving spatial resolution and/or SNR of images using super-resolution are known.

FIG. **6** is a process flow diagram that illustrates a first embodiment method **600** for obtaining images using an imaging system that includes an x-ray source and an x-ray detector. The imaging system may be similar or identical to the systems shown in FIGS. **1**-**5**C. The detector may be a "hybrid" CT/fluoroscopy detector such as shown in FIGS. **5**A-**5**C where the x-ray beam is collimated to expose the second portion **503** of the detector area, as shown in FIG. **5**C. Alternately, the detector may be a flat panel detector for fluoroscopy/cone beam imaging. The detector may have a size and shape that corresponds to the second portion **503** of the detector shown in FIGS. **5**A-**5**C (i.e., with the elongated "wing" sections of the first portion **502** omitted), and may include, for example, a GOS scintillator, 2D anti-scatter plates, a curved detector surface, etc. Alternately, the detector may be a conventional flat panel detector with a flat, planar detector surface and may include, for example, a cesium iodide scintillator material. In some embodiments, the detector may be included in an external beam radiation treatment system, such as a linear accelerator (LINAC) system, as described in further detail below.

In block **601**, a plurality of first x-ray images of an object (e.g., a human or animal patient) may be obtained at a detector with a first spatial resolution while the detector is moved slightly with respect to the source between images. The magnitude of the movement of the detector with respect to the source may be a sub-pixel length (i.e., less than the center-to-center spacing, Sp, between adjacent pixels).

In block **603**, at least one second image of the object may be generated using the plurality of first x-ray images, where the at least one second image has a spatial resolution that is different than the first spatial resolution. The plurality of first images may be used to generate the at least one second image using a super-resolution technique as described above. In particular, information from the plurality of first images may be intelligently merged to produce a second super-resolution (SR) image which may have more information than is contained in any one of the original images. The SR image may be upsized (i.e., contain more pixels) than the original images(s) and have an improved spatial resolution as described above. The SR image may also have an improved SNR compared to the original image(s).

The at least one second image may be generated by a suitable computing device having a memory and a processor coupled to the memory, such as the computer **46** shown in FIG. **3**. Alternately, the at least one second image may be generated using a memory and processor located on or within the detector. The processor may be configured with processor-executable instructions to receive electronic representations of the first images (i.e., frames of x-ray image data) obtained by the detector and to use the first images to generate a second image having improved spatial resolution and SNR, for example, using a super-resolution algorithm. In embodiments, as each first image is obtained and read out from the detector, the image (e.g., image frame) may be stored in a memory, and then combined with data from additional image(s) (i.e., preceding and/or subsequent frames from the detector) to generate a super-resolution image having an improved spatial resolution. For example, each second image may be generated by combining data from a certain number (e.g., 2-10, such as 3-5) of image frames from the detector. The data may be combined using a super-resolution technique, as described above.

In one exemplary embodiment, the detector may read out image frames at a rate that is greater than 30 Hz, such as at least about 60 Hz (e.g., 60-300 Hz), while the detector is moved a sub-pixel amount with respect to the x-ray source. The processor may combine a certain number of image frames (e.g., 5 frames) to generate a SR image. For example, if the detector system is reading out image frames at a first frame rate (e.g., 300 Hz), the processor may combine a set number of image frames (e.g., 5 frames) to produce SR images at a lower frame rate (e.g., 60 Hz). Alternately, the processor may generate SR images at the same frame rate as the image data is acquired by the detector by combining a set number of preceding frames with each new frame read out from the detector.

The generated SR images may be sent to a display device (e.g., workstation computer, monitor device, etc.) for real-time display of the SR images. In embodiments, the SR images may be transmitted via a wireless communication link from the gantry **40** to an external computing device. In some embodiments, a plurality of SR images may be generated from different projection angles as the rotor **41** rotates to different positions around the gantry **40**. The plurality of SR images may be used as an input to a tomographic reconstruction algorithm (e.g., a backprojection algorithm), and to generate a tomographic reconstruction (e.g., cone-beam CT reconstruction) of the object using the SR images.

The imaging system may further include an apparatus for moving the detector relative to the x-ray source while the plurality of first images are obtained by the detector. For example, one or more actuators may be coupled to the detector system (e.g., to the detector chassis) to drive the motion of the detector relative to the x-ray source. FIG. **5**C schematically illustrates an actuator device **507** that is configured to controllably displace the detector **501** by a sub-pixel amount. The actuator device **507** may comprise one or more piezoelectric motors, for example. The actuator device **507** may displace the detector **501** in at least one direction (i.e., in the x-axis direction **508** and/or the z-axis direction **509**) and preferably in two orthogonal directions. The magnitude and direction of the displacement of the detector **501** between each image obtained by the detector may be known (e.g., via encoder feedback) or may be estimated or derived based on the operating conditions of the actuator device **507**. Thus, complicated motion estimation processes that are typically used in super-resolution techniques may be greatly simplified or eliminated.

If the displacement of the detector relative to the source between multiple x-ray images is precisely known (e.g., via encoder data), then the generated SR images may exhibit a 4-5 times improvement in spatial resolution relative to the originally-obtained images. However, even where the displacement is not known, a motion estimation algorithm may be used to improve the spatial resolution by at least about a factor of two.

Super-resolution techniques may be used to improve spatial resolution and also to increase SNR of the originally obtained image. Thus, in embodiments, the plurality of first images may be obtained using a relatively lower output power (i.e., lower x-ray dosing) from the x-ray source **43**. This may decrease the SNR of the first images obtained at the detector. However, the SNR may be restored in the at least one second image generated from the plurality of lower SNR first images using a super-resolution technique as described above. In some embodiments, the SNR of the first plurality of images may be below generally acceptable levels for use in diagnostic imaging. However, the SNR of the generated second image(s) may be sufficiently high for use in diagnostic imaging.

Further embodiments include methods and systems for performing cone beam CT imaging using scan paths that follow non-planar trajectories. In a typical cone beam CT scan, the x-ray source and detector rotate around the patient in a single scan plane along a circular trajectory. A plurality of 2D x-ray projections are obtained by the detector from different projection angles as the source and detector rotate, and are used to generate a 3D reconstruction of a region of interest. In the tomographic reconstruction, the highest image quality is found in the central (axial) slice of the reconstruction, with image quality decreasing as a function of distance from the central slice. This is because the central slice, which corresponds to the scan plane of the x-ray source, has the most complete set of projection data from every projection angle, while there is less projection data for the slices further away from the central slice. This may result in artifacts in the reconstructed image, particularly in the peripheral slices.

In various embodiments, a cone beam CT imaging system that uses non-planar scan trajectories, such as a sinusoidal trajectory, may be utilized to obtain more complete set of projection data for reconstruction of slices outside of the central slice. The resulting 3D reconstruction may be characterized by improved image quality and reduced artifacts. The imaging system may be similar or identical to any of the systems shown in FIGS. **1**-**5**C. The detector may be a panel detector having a length dimension of at least about 20 cm (e.g., 30-40 cm) and a width dimension of at least about 20 cm (e.g., 30-40 cm). The detector may be a "hybrid" CT/fluoroscopy detector such as shown in FIGS. **5**A-**5**C where the x-ray beam is collimated to expose the second portion **503** of the detector area. In some embodiments, the imaging system may be integrated in an external beam radiation treatment system, such as a linear accelerator (LINAC) system, as described in further detail below.

FIGS. **7**A-**7**G schematically illustrate a method of performing cone beam CT imaging of an object (e.g., a human or animal patient) using a sinusoidal circular trajectory. The imaging system **700** in this embodiment is similar to the imaging system shown in FIG. **1**, and includes, for example, a base **20**, a patient support **60** supported above the base, and an imaging gantry **40** supported by a gimbal **30**, where a drive mechanism **70** drives the translation of the gantry **40** and gimbal **30** along the length of the base **20**. The system controller **113** may control the imaging system **700** to perform a coordinated translation of the gantry **40** as the x-ray source rotates around the patient **701** (e.g., via a rotor **41** within the gantry **40** as shown in FIG. **3**) so that the focal spot of the x-ray source follows a sinusoidal-shaped trajectory as it rotates around the patient **701**. This is schematically illustrated in FIGS. **7**A-**7**G, which shows the focal spot of the x-ray source (indicated by point, P) rotating 180° from the top of the gantry **40** (FIG. **7**A) to the bottom of the gantry (FIG. **7**G). As the source rotates, the system controller **113** controls the drive system **70** to translate the gantry **40** back and forth along the z-axis so that the focal spot, P, follows a sinusoidal path around the patient **701**, as indicated by the dashed line in FIGS. **7**B-**7**G. As the gantry **40** continues to rotate around the patient **701**, the gantry may continue to translate along the z-axis. In the embodiment of FIGS. **7**A-**7**G, for each 360° rotation of the source around the patient, the trajectory of the x-ray focal spot, P, may trace out three "peaks" and three "valleys" in the z-axis direction, or a sinusoid of order m=3. In other embodiments, the sinusoid may be of order m=2, m=4, m=5, etc.

The displacement of the gantry **40** (i.e., ±d shown in FIGS. **7**A-**7**G) may be less than the width of the detector in the z-direction. In embodiments, the gantry **40** may be displaced by a fraction of the detector width, such as ½, ⅓, ¼, ⅙, ⅛, etc. of the detector width. The positions of the gantry **40** along the z-axis during the scan may be provided to the CT reconstructor (e.g., via encoder data) and combined with the projection data and rotor encoder data to perform tomographic reconstruction.

An imaging system such as shown in FIG. **2** may also be used to perform cone beam CT imaging using a sinusoidal circular trajectory. For example, for a patient supported in a horizontal configuration, drive mechanism **208** may be controlled to translate the gantry **40** and support column **201** back-and-forth in a horizontal direction in coordination with the rotation of the x-ray source so that the source follows a sinusoidal trajectory relative to the patient. For a patient supported in a vertical, weight-bearing configuration, drive mechanism **203** may be controlled to translate the gantry **40** up-and-down on the support column **201** to provide a sinusoidal scan trajectory. When the patient is supported in a tilted configuration, drive mechanisms **203** and **208** may be controlled to move the gantry **40** up and down along the tilted axis to provide a sinusoidal scan trajectory.

In a sinusoidal circular trajectory as shown in FIGS. **7**A-**7**G, the x-ray source rotates around the patient while the gantry translates along the patient axis to provide a sinusoidal scan trajectory over the surface of an imaginary cylinder. In further embodiments, the x-ray source may rotate around the patient in coordination with a rotation of the gantry about its isocenter to provide a sinusoidal spherical trajectory. In a spherical circular trajectory, the sinusoidal-shaped path of the x-ray focal spot extends over the surface of an imaginary sphere centered on the isocenter of the gantry.

In the imaging system of FIG. **1**, for example, the system controller **113** may control the motorized system **112** to perform a coordinated rotation of the gantry **40** about axis **102** (i.e., "wag" rotation) as the x-ray source rotates around the patient **701** so that the focal spot of the x-ray source follows a sinusoidal spherical trajectory. This is schematically illustrated in FIGS. **8**A and **8**B, which illustrates the "wag" rotation of the gantry **40**. Alternately or in addition, the system controller **113** may control the motorized system **115** (see FIG. **1**) to perform a coordinated rotation of the gantry **40** about axis **107** (i.e., "tilt" rotation) to provide a sinusoidal spherical scan trajectory.

Similarly, in the imaging system of FIG. **2**, the system controller **213** may control drive mechanisms **212** and **208** to rotate the gantry **40** about pivot axis **211** while translating the gantry **40** in the direction opposite to the direction of rotation to provide isocentric "wag" rotation while the x-ray source rotates to provide a sinusoidal spherical scan trajectory. Alternately or in addition, the system controller **213** may control drive mechanism **205** to perform a coordinated rotation of the gantry **40** about axis **207** with the rotation of the x-ray source to provide a sinusoidal spherical scan trajectory.

FIG. **9** illustrates an external beam radiation treatment system **900**, such as a linear accelerator (LINAC) system, a gamma knife system or a proton beam radiosurgery system, that includes an integrated x-ray imaging system **901**, in accordance with various embodiments. FIG. **9** illustrates a LINAC system that includes a rotating gantry **902** and a patient support **905** (e.g., couch) on which a patient may be supported. The gantry **902** may rotate relative to a support structure **912** in the direction of arrow **913**. The gantry **902** includes a head **903** that emits a high-energy (e.g., MeV-range) beam of radiation that may be directed to a target location (e.g., a tumor) within the patient's body. The head **903** may be rotated around the patient to irradiate the patient from any angle. The patient support **905** may be moveable to adjust the position and/or orientation of the patient relative to the head **903**.

In a typical external beam radiation treatment, one or more pre-operative images (e.g., CT scan and/or MRI images) of the patient may be used for pre-treatment planning. The pre-treatment planning may include identifying the location of the target tissue (e.g., tumor(s)) to be irradiated, as well as identifying the location of other tissue (e.g., "organs at risk") to be avoided. At the time of treatment, an imaging device located in the treatment suite may be used to obtain additional images of the patient, such as 2D x-ray fluoroscopic images and/or a 3D x-ray cone beam CT image. The additional image(s) may be matched to the pre-operative images to enable the patient to be positioned in the same coordinate system (relative to the LINAC head **903**) as was used for the pre-operative planning. In the embodiment shown in FIG. **9**, the imaging device **901** may be integrated with the treatment system **900**, and may include an x-ray source **906** and a detector **907** mounted opposite one another on the rotating gantry **902**. The source **906** and detector **907** may each be offset from the head **903** by 90° and may rotate together with the head **903** on the gantry **902**. The source **906** and detector **907** may be mounted to the gantry **902** by a pair of arms **908**, **909** (e.g., robotic arms) that may enable a limited amount of controlled movement of the source **906** and detector **907** relative to the gantry **902** and head **903**.

External beam radiation treatment systems that include integrated x-ray imaging equipment commonly use flat panel detectors. The image quality obtainable from such detectors is generally not sufficient for use directly for pre-treatment planning purposes. The images from such detectors are typically used to match bony landmarks visible in the images to the same landmarks in pre-operative images used in pre-treatment planning, which are generally of much higher image quality and may enable clearer visualization of tumors and other soft tissue of the patient. A patient may receive external beam radiation treatment therapy over the course of multiple (e.g., daily) treatment sessions that can last several weeks. During the course of the therapy sessions, the target tissue (e.g., tumor) can shrink significantly and can also move within the patient's body. However, changes in the size, shape and/or location of the target tissue may not be evident from the images (e.g., 2D fluoroscopy and/or cone beam CT images) obtained using the x-ray imaging equipment located at the point-of-treatment.

Various embodiments of an external beam radiation treatment system **900** having an integrated x-ray imaging system **901** utilize a detector **907** that provides improved image quality relative to a conventional flat panel detector. In embodiments, the detector **907** may be similar to the detectors described above with reference to FIGS. **4**A-**4**G and **5**A-**5**C. For example, the detector **907** may include a curved profile in which the detector elements or modules are arranged along an arc centered on the focal spot of the x-ray source **906**. The detector **907** may have a curved profile along the length dimension and/or along the width dimension. The detector **907** may be a modular detector such as described with reference to FIGS. **4**A-**4**G. Each module may include a scintillator (e.g., GOS), a photodiode array, and an electronics assembly including analog-to-digital (A/D) converter circuitry for converting the output signals from the photodiode array to digital signals. A plurality of modules may be mounted to a chassis in abutting fashion to form a detector panel having a desired size, shape and cross-sectional profile. The detector **907** may utilize a double- or multiple-buffering configuration and may enable continuous exposure, as described above. The detector **907** may also include a 1D or 2D anti-scatter assembly, as is also described above. The detector **907** according to various embodiments may provide diagnostic-quality images at the treatment site that may be used for pre-treatment planning or modification to an existing treatment plan for external beam radiation therapy of a patient.

The imaging system **901** may also be used to generate images having improved spatial resolution and/or SNR using a super-resolution (SR) technique, as described above. In particular, a plurality of first x-ray images of the patient may be obtained using the detector **907** while the detector **907** is moved slightly with respect to the source **906** between images. The magnitude of the detector movement **907** may be in the sub-pixel range (i.e., less than the center-to-center spacing, Sp, between adjacent pixels). The first images may be fed to a computer **910** that may use the first image to generate at least one second image using a super-resolution technique as described above. The generated second image may be upsized (i.e., contain more pixels) than the first images and may have an improved spatial resolution compared to the first images. The second image may also have an improved SNR compared to the first images.

The movement of the detector **907** relative to the source **906** between successive x-ray images may be achieved by moving the arm **909** to which the detector **907** is mounted. The system **900** may include a controller **911** (e.g., a computer) that may be configured to send control signals the arms **908**, **909** to control the movements of the arms **908**, **909**. The controller **911** may control the arm **909** to move the detector **907** by a sub-pixel amount between the acquisition of each of the first images that are used to generate the super-resolution images. Alternately or in addition, a separate actuator system (e.g., one or more piezoelectric motors) may be operatively coupled to the detector **907** to drive the motion of the detector **907** relative to the x-ray source **906**, as described above.

In various embodiments of a system **900** as shown in FIG. **9**, the gantry **902** may be rotated to obtain 2D x-ray images of the patient from desired projection angles. Alternately or in addition, the gantry **902** may rotate around the patient while the x-ray source **906** and detector **907** obtain a set of projection data that may be used to generate a 3D cone-beam CT reconstruction. The detector **907** size may provide a limit on the field-of-view of the images that may be obtained. For example, a detector **907** with a dimension of 30-40 cm may only provide a reconstruction volume of ~15-20 cm diameter in the axial plane of the patient.

Various embodiments include controlling the arm **909** to move the detector **907** along an arc or line with respect to the focal spot of the x-ray source to effectively increase the field-of-view of the detector **907**. This is schematically illustrated in FIG. **10**, which shows the detector **907** (in phantom) moved to different positions along the directions of arrows **1001**, **1002**. The controller **911** may control the arm **909** to move the detector **907** by a desired distance (e.g., a fraction of a panel length, such as ½ a panel length, a full panel length or more) from an initial position. The system **900** may obtain multiple images from the same projection angle with the detector translated to different positions with respect to the x-ray focal spot. The multiple images may be combined in software (e.g., using computer **910**) to generate images having a larger effective field-of-view. To perform an imaging scan, the gantry **902** may rotate the source **906** and detector **907** around the patient to obtain a first set of projection data with the detector **907** at a first position relative to the x-ray focal spot, the arm **909** may move the detector **907** to a second position relative to the focal spot, and the gantry **902** may rotate the source **906** and detector **907** around the patient again to obtain a second set of projection data with the detector **907** located in the second position. The process may optionally be repeated with the detector **907** moved to additional positions relative to the x-ray focal spot. In one example, the gantry **902** may perform a full 360° rotation around the patient to obtain a first set of projection data at a plurality of rotational positions, the detector **907** may then be moved (e.g., via a half- or full-panel length) and the gantry **902** may perform another 360° rotation (e.g., a counter-rotation) to obtain a second set of projection data at the same rotational positions. Alternately, the gantry **902** may perform a partial rotation (e.g., a 180° rotation) with the detector **907** in a first position, the detector **907** may be moved to a second position, and the gantry **902** may continue the rotation (e.g., another 180°).

In other embodiments, an imaging scan may be performed by rotating the gantry **902** to a plurality of rotation angles around the patient, and at each rotation angle, moving the detector **907** to different position(s) relative to the focal spot and obtaining multiple sets x-ray image data with the detector **907** moved to each position.

In embodiments, the x-ray source **906** may include a mechanism that alters a characteristic of the output x-ray beam based on the position of the detector **907**. In one embodiment, the x-ray source **906** may include an adjustable collimator that may shape the beam output such that the detector **907** remains exposed to x-rays at it moves to different positions, while portion(s) of the beam not incident on the detector **907** are blocked. Alternately or in addition, the arm **908** may be controlled to pivot the x-ray source **906** about the stationary focal spot (indicated by arrow **1003** in FIG. **10**) in coordination with the movement of the detector **907** so that the x-ray beam remains centered on detector **907**. This may minimize the patient's exposure to x-rays in regions outside of the field-of-view of the detector and enable safer and more efficient dose utilization.

In further embodiments, the system **900** may utilize non-planar scan trajectories, such as a sinusoidal trajectory, to obtain a more complete set of projection data for cone-beam CT imaging. In one example, the controller **911** may control the arms **908**, **909** to translate the source **906** and detector **907** back and forth along the length of the patient in the direction of arrows **1101**, **1102**, as shown in FIG. **11**. The translation of the source **906** and detector **907** may be coordinated with the rotation of the gantry **902** so that the focal spot of the x-ray source follows a sinusoidal circular trajectory as it rotates around the patient. Alternately, the arms **908**, **908** may be controlled to move the source **906** and detector **907** over an imaginary spherical surface to provide a sinusoidal spherical trajectory, as described above.

In embodiments, the controller **911** may control the arms **908**, **909** to move the source **906** and detector **907** in a sinusoidal scan trajectory while arm **909** may move the detector **907** along an arc or line with respect to the focal spot of the x-ray source **906** to effectively increase the field-of-view of the detector **907**. For example, the source **906** and detector **907** may move back and forth in the z-axis direction as shown in FIG. **11** to provide a sinusoidal circular scan trajectory and the detector **907** may also be moved in the transverse direction with respect to the focal spot of the source **906** to extend the effective field-of-view, as illustrated in FIG. **10**. For example, the gantry **902** may perform a full (i.e., 360°) or partial (e.g., 180°) rotation with the source **906** and detector **907** moving in a sinusoidal trajectory and the detector **907** in a first position with respect to the x-ray focal spot. The detector **907** may then be moved to a second position with respect to the focal spot, and the gantry **902** may perform another rotation or continue the current rotation around the patient with the source **906** and detector **907** moving in a sinusoidal trajectory.

Translating the source **906** and/or detector **907** on the arms **908**, **909** as schematically shown in FIG. **11** may enable the field-of-view of the imaging system **901** to be effectively increased in the z-axis dimension (i.e., along the patient length). This may enable the system to follow a contrast agent injected into the patient as it progresses through the patient's body, for example. The translation of the source **906** and detector **907** may also enable the imaging system **901** to perform helical cone-beam CT scans along the length of the patient.

In a typical helical CT scan, the source and detector are rotated continuously around the patient while either the patient is moved within the imaging bore or the imaging gantry is moved over the patient so that the x-ray source defines a helical or spiral trajectory around the patient. However, many rotating-gantry external beam radiation systems (e.g., LINAC systems) can rotate and counter-rotate over a limited range but do not enable continuous rotation of the gantry. This may limit the field-of-view of the CT reconstruction along the length of the patient.

Further embodiments include operating an external beam radiation system **900** having an integrated imaging system **901** to perform a reverse helical scan of a patient. In embodiments, the controller **911** may control the arms **908**, **909** to translate the source **906** and the detector **907** in a first translation direction along the length of the patient as the rotor rotates the source **906** and detector **907** in a first rotational direction (i.e., clockwise or counter-clockwise) around the patient. The gantry **902** may perform a full (i.e., 360°) or partial (e.g., 180°, 270°, etc.) rotation around the patient. The gantry **902** may then counter-rotate in a second rotational direction opposite the first rotational direction while the arms **908**, **909** may continue to translate in the first translation direction. The gantry **902** may continue to alternate its rotational direction while the arms **908**, **909** continue to translate the source **906** and detector **907** in the first direction along the patient length to provide a reverse helical scan trajectory. This is schematically illustrated by FIG. **12**, which shows the trajectory (indicated by arrow **1201**) of the detector **907** around and along the length of the patient. In embodiments, the source **906** and detector **907** may translate at least about 0.5 meters along the length of the patient during a reverse helical scan. This may enable the CT reconstruction to have a larger field-of-view along the patient length, which may encompass, for example, complex tumors that extend down the length of the patient's spine.

In some embodiments, the system **900** may perform a scan along a reverse helical trajectory while arm **909** may move the detector **907** along an arc or line with respect to the focal spot of the x-ray source **906** to effectively increase the field-of-view of the detector **907**. For example, the source **906** and detector **907** may translate in a first direction while the gantry rotates and counter-rotates as described above while the detector **907** is in a first position relative to the focal spot of the detector **907**. The arm **909** may then move the detector **907** to a second position relative to the source **906** (e.g., by translating the detector **907** by one-half panel width), and the system **900** may perform another reverse helical scan in the reverse direction by translating the source **906** and detector **907** in a second direction back down the length of the patient while the gantry rotates and counter-rotates.

In some embodiments, the detector **907** may have an integrated processor unit that may be configured to perform all or a portion of the image processing operations, including tomographic reconstruction. The integrated processor unit may be an alternative or in addition to the separate computer **910** shown in FIGS. **9-12**. The integrated processor unit may receive inputs of encoder position(s) (e.g., indicating the rotation position of the gantry **902** and/or the configurations of the arms **908**, **909**), source-to-detector distance, x-ray photon flux and/or x-ray source temperature (e.g., from a reference detector as described in U.S. Pat. No. 9,111,379). The integrated processor unit may be configured to perform various image correction techniques on the image data, such as offset correction, gain correction and/or pixel correction. In some embodiments, the integrated processor unit may also perform other real-time processing operations (e.g., for 2D fluoroscopy), including edge enhancement, recursive noise reduction and super resolution techniques. The processed images may be transmitted from the detector **907** to another entity, such as to a separate workstation.

Although the external beam radiation treatment system **900** shown in FIGS. **9-12** includes a pair of arms **908** and **909** extending from opposite sides of the gantry **902**, it will be understood that other configurations are possible. For example, a single arm (e.g., **908** or **909**) may extend from the gantry **902** and support both the source **906** and detector **907**. A second arm may extend in a transverse direction (e.g., beneath the patient table **905** in FIG. **9**) to couple the source **906** and detector **907** and maintain a constant spacing between the source **906** and detector **907**.

FIG. **13** illustrates yet another embodiment of an imaging system **1300**. In this embodiment, the imaging system **1300** includes an O-shaped imaging gantry **1340** that is mounted to a support structure **1301** in a cantilevered fashion. The imaging system **1300** may be an x-ray imaging system that may be used to obtain 2D fluoroscopic images and/or 3D tomographic images of an object located within the bore **1316** of the gantry. At least one of an x-ray source **1302** and an x-ray detector **1303** may rotate around the interior of the gantry (as shown by arrow **1308**) to obtain images of an object within the bore **1316** from different projection angles. The support structure **1301** may comprise a mobile cart **1306** that is attached to one side of the gantry via an attachment mechanism **1307**. The attachment mechanism **1307** may include one or more motorized systems that enable the gantry **1340** to translate and/or rotate with respect to at least a portion of the cart **1306**. For example, in embodiments the gantry **1340** may be raised or lowered relative to the cart **1306** in the direction of arrow **1309**. The gantry **1340** may also translate relative to the cart **1306** over a limited range (e.g., 30-50 cm, such as about 40 cm) in the direction of the z-axis **1310** (i.e., into and out of the page in FIG. **13**). In addition, in some embodiments the gantry **1340** may be rotated with respect to the cart **1306** along at least one rotational axis. For example, the gantry **1340** may be tilted with respect to the cart **1306** about a horizontal axis **1311**. The gantry **1340** may also be pivoted with respect to the cart **1306** about a vertical axis **1312**. The gantry **1340** may perform an isocentric "wag" rotation via a combination of pivot of the gantry **1340** about the vertical axis **1312** and a translation of the gantry **1340** along the z-axis **1310** with respect to the cart **1306**. A control system **1313** (e.g., a computer) may control the operation of the imaging system **1300**, including the translational and rotational movements of the gantry **1340** relative to the cart **1306**.

In some embodiments, the detector **1303** in the system **1300** shown in FIG. **13** may be a conventional flat-panel detector. Alternately, the detector may be a diagnostic quality CT detector similar to the detectors described above with reference to FIGS. **4**A-**4**G, **5**A-**5**C and **9**-**12**. For example, the detector **1303** may include a curved profile and/or an anti-scatter assembly located over the detector elements. The detector **1303** may utilize a high-speed scintillator (e.g., GOS) and associated electronics, and may enable continuous exposure with a high readout rate. In some embodiments, the detector **1303** may be configured to provide improved spatial resolution and/or SNR using super resolution techniques, as described above.

In embodiments, the control system **1313** may control the imaging system **1300** to translate and/or rotate the gantry **1340** in coordination with the rotation of the source **1302** and detector **1303** to perform an imaging scan using a sinusoidal scan trajectory, such as a sinusoidal circular trajectory or a sinusoidal spherical trajectory, as discussed above. In some embodiments, the control system **1313** may control the imaging system **1300** to translate the gantry **1340** along the z-axis in coordination with a rotation and counter-rotation of the source **1302** and detector **1303** within the gantry **1340** to perform a cone-beam CT scan using a reverse helix trajectory. In some embodiments, the detector **1303** may include a mechanism that moves the detector **1303** by a sub-pixel amount to enable super-resolution imaging as described above.

FIG. **14** is a system block diagram of a computing device **1400** useful for performing and implementing the various embodiments described above. The computing device **1300** may perform the functions of a control system **113**, **213**, **910**, **1313** for an imaging system and/or a system for processing x-ray image data. While the computing device **1400** is illustrated as a laptop computer, a computing device providing the functional capabilities of the computer device **1400** may be implemented as a workstation computer, an embedded computer, a desktop computer, a server computer or a handheld computer (e.g., tablet, a smartphone, etc.). A typical computing device **1400** may include a processor **1401** coupled to an electronic display **1404**, a speaker **1406** and a memory **1402**, which may be a volatile memory as well as a nonvolatile memory (e.g., a disk drive). When implemented as a laptop computer or desktop computer, the computing device **1400** may also include a floppy disc drive, compact disc (CD) or DVD disc drive coupled to the processor **1401**. The computing device **1400** may include an antenna **1410**, a multimedia receiver **1412**, a transceiver **1418** and/or communications circuitry coupled to the processor **1401** for sending and receiving electromagnetic radiation, connecting to a wireless data link, and receiving data. Additionally, the computing device **1400** may include network access ports **1424** coupled to the processor **1401** for establishing data connections with a network (e.g., LAN coupled to a service provider network, etc.). A laptop computer or desktop computer **1400** typically also includes a keyboard **1414** and a mouse pad **1416** for receiving user inputs.

The foregoing method descriptions are provided merely as illustrative examples and are not intended to require or imply that the steps of the various embodiments must be performed in the order presented. As will be appreciated by one of skill in the art the order of steps in the foregoing embodiments may be performed in any order. Words such as "thereafter," "then," "next," etc. are not necessarily intended to limit the order of the steps; these words may be used to guide the reader through the description of the methods. Further, any reference to claim elements in the singular, for example, using the articles "a," "an" or "the" is not to be construed as limiting the element to the singular.

The various illustrative logical blocks, modules, circuits, and algorithm steps described in connection with the embodiments disclosed herein may be implemented as electronic hardware, computer software, or combinations of both. To clearly illustrate this interchangeability of hardware and software, various illustrative components, blocks, modules, circuits, and steps have been described above generally in terms of their functionality. Whether such functionality is implemented as hardware or software depends upon the particular application and design constraints imposed on the overall system. Skilled artisans may implement the described functionality in varying ways for each particular application, but such implementation decisions should not be interpreted as causing a departure from the scope of the present invention.

The hardware used to implement the various illustrative logics, logical blocks, modules, and circuits described in connection with the aspects disclosed herein may be implemented or performed with a general purpose processor, a digital signal processor (DSP), an application specific integrated circuit (ASIC), a field programmable gate array (FPGA) or other programmable logic device, discrete gate or transistor logic, discrete hardware components, or any combination thereof designed to perform the functions described herein. A general-purpose processor may be a microprocessor, but, in the alternative, the processor may be any conventional processor, controller, microcontroller, or state machine. A processor may also be implemented as a combination of computing devices, e.g., a combination of a DSP and a microprocessor, a plurality of microprocessors, one or more microprocessors in conjunction with a DSP core, or any other such configuration. Alternatively, some steps or methods may be performed by circuitry that is specific to a given function.

In one or more exemplary aspects, the functions described may be implemented in hardware, software, firmware, or any combination thereof. If implemented in software, the functions may be stored on as one or more instructions or code on a non-transitory computer-readable medium. The steps of a method or algorithm disclosed herein may be embodied in a processor-executable software module executed which may reside on a non-transitory computer-readable medium. Non-transitory computer-readable media includes computer storage media that facilitates transfer of a computer program from one place to another. A storage media may be any available media that may be accessed by a computer. By way of example, and not limitation, such non-transitory computer-readable storage media may comprise RAM, ROM, EEPROM, CD-ROM or other optical disk storage, magnetic disk storage or other magnetic storage devices, or any other medium that may be used to carry or store desired program code in the form of instructions or data structures and that may be accessed by a computer. Disk and disc, as used herein, includes compact disc (CD), laser disc, optical disc, digital versatile disc (DVD), floppy disk, and blu-ray disc where disks usually reproduce data magnetically, while discs reproduce data optically with lasers. Combinations of the above should also be included within the scope of non-transitory computer-readable storage media. Additionally, the operations of a method or algorithm may reside as one or any combination or set of codes and/or instructions on a machine readable medium and/or computer-readable medium, which may be incorporated into a computer program product.

The preceding description of the disclosed aspects is provided to enable any person skilled in the art to make or use the present invention. Various modifications to these aspects will be readily apparent to those skilled in the art, and the generic principles defined herein may be applied to other aspects without departing from the scope of the invention. Thus, the present invention is not intended to be limited to the aspects shown herein but is to be accorded the widest scope consistent with the principles and novel features disclosed herein.

What is claimed is:

1. An x-ray imaging system, comprising:

an O-shaped gantry including a housing and defining an imaging bore;

an x-ray source located within the housing of the gantry;

a detector system located within the housing of the gantry opposite the x-ray source, the detector system including a plurality of x-ray sensitive detector elements defining a contiguous detector area having an elongated central portion and a pair of peripheral portions arranged on opposing sides of the central portion to define a panel region; and

a drive system for rotating the x-ray source and the detector system around the imaging bore;

wherein the central portion defines a central portion length and a central portion width, and the panel region defines a panel length and a panel width, the central portion length being greater than the panel length, and the panel width being greater than the central portion width.

2. The x-ray imaging system of claim **1**, wherein the x-ray source includes an adjustable collimator configured to direct a beam of x-ray radiation onto the central portion of the detector area to perform a fan-beam computed tomography (CT) imaging scan.

3. The x-ray imaging system of claim **2**, wherein the adjustable collimator is further configured to direct a beam of x-ray radiation onto to the panel region of the detector area to perform 2D x-ray fluoroscopy and/or cone-beam CT imaging.

4. The x-ray imaging system of claim **3**, wherein the central portion of the detector area is configured to collect at least 32 slices of x-ray data simultaneously as the x-ray source and detector system rotate within the gantry during a fan-beam computed tomography (CT) imaging scan.

5. The x-ray imaging system of claim **1**, wherein the detector elements include:

a layer of scintillator material;

a photodiode array optically coupled to the layer of scintillator material; and

an electronics assembly coupled directly behind the photodiode array and including an analog-to-digital (A/D) converter for converting output signals from the photodiode array to digital signals.

6. The x-ray imaging system of claim **5**, wherein the layer of scintillator material includes gadolinium oxysulfide (GOS).

7. The x-ray imaging system of claim **5**, wherein the detector system is configured to read-out a plurality of frames of image data while the detector elements are continuously exposed to x-ray radiation from the x-ray source.

8. The x-ray imaging system of claim **1**, wherein adjacent x-ray sensitive detector elements are spaced from each other at a common spacing distance.

9. The x-ray imaging system of claim **8**, wherein a spacing ratio of the central portion length to the spacing distance is greater than 250:1.

10. The x-ray imaging system of claim **8**, wherein the spacing distance is less than 2 millimeters.

11. The x-ray imaging system of claim **1**, further comprising an anti-scatter apparatus located over the contiguous detector area.

12. The x-ray imaging system of claim **1**, wherein the detector system defines a detector surface that is curved or angled along its length and defines a semicircular arc centered on a focal spot of the x-ray source.

13. The x-ray imaging system of claim **12**, wherein the detector surface is further curved or angled along its width and defines a semicircular arc centered on the focal spot of the x-ray source.

14. The x-ray imaging system of claim **1**, wherein the detector system includes a multiple buffering configuration configured such that at least one buffer reads-out previously-collected digital image data while new digital image data accumulates in another buffer.

15. The x-ray imaging system of claim **1**, further comprising an apparatus for moving the detector elements relative to the x-ray source by a sub-pixel amount.

16. The x-ray imaging system of claim **15**, further comprising a processing device coupled to the detector system and being configured with processor-executable instructions to perform operations including:

receiving, from the detector system, a plurality of first x-ray images of an object with a first spatial resolution that are obtained while the detector elements are moved relative to the x-ray source; and

generating at least one second image of the object using the plurality of first x-ray images, wherein the at least one second image is a super resolution (SR) image that has a greater spatial resolution and/or signal-to-noise (SNR) ratio compared to the first x-ray images.

17. The x-ray imaging system of claim **16**, wherein a width ratio of the panel width to the central portion width is at least 1.2:1.

18. The x-ray imaging system of claim **1**, wherein a length ratio of the central portion length to the panel length is at least 1.25:1.

19. The x-ray imaging system of claim **1**, wherein a length ratio of the central portion length to the panel length is at least 2:1.

20. The x-ray imaging system of claim **19**, wherein a width ratio of the panel width to the central portion width is at least 1.2:1.

21. The x-ray imaging system of claim **1**, wherein the central portion length is at least 1 meter.

22. The x-ray imaging system of claim **21**, wherein the central portion length is less than 0.5 meters; and

wherein the central portion width is greater than 0.3 meters.

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